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Effects of Slips and Trips on Resultant Lumbar Kinematics, Lumbar Muscle Activity and Low-Back Loads

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By

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Approved by:

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Abstract

Slips, trips, and falls (STFs) represent one of the leading causes of occupational injuries and fatalities. In particular, many prior reports have linked STFs with the onset of low-back disorders, which, depending on the severity of the incident, can leave the worker physically limited both in the workplace and at home. In contrast, the incidence and outcomes of loads acting on the low back due to a slip and trip *that does not lead to a fall* (i.e., slip/trip without fall: STWF) remain only marginally investigated to date. To address this research deficit, this quantitative study was designed to explore selected physiological outcomes of STWFs. In terms of methodology, participants completed several walking trials during which two unexpected perturbations involving a slip and trip were introduced (a harness prevented a fall). A biomechanical model developed using the AnyBody modeling software yielded trunk kinematics and muscle geometry. These outputs - along with the electromyography of fourteen lumbar flexor and extensor muscles - were employed as input data for our 3D, dynamic, EMG-based lumbar spine model. Results of (a) lumbar kinematics (range of the motion of the trunk relative to the pelvis), (b) lumbar muscle activity, (c) lumbosacral reaction forces, and (d) moments all indicated more than a two-fold increase during the slip and trip trials compared to normal walking. Specifically, reported values for the slip trial were (a) 45°, (b) 0.694, (c) 2939 N, and (d) 52 Nm; Reported values for the trip trial were (a) 42°, (b) 0.691, (c) 2898 N, and (d) 50 Nm; and the analogous figures for normal walking were (a) 19°,(b) 0.195, (c) 1174 N, and (d) 16 Nm. Findings from this study can be used to develop interventions to avoid such incidents; for example, to determine specific training parameters (e.g., frequency, duration, and intensity) to optimize a developed intervention's effectiveness. Such approaches may lead to the control of specific mechanisms involved with low-back disorders consequent to a slip or trip, and potentially reduce the risk for slip- and trip-related injuries.

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1. Significance

1.1. Introduction and Background

Slip and fall incidents represent a major threat to the safety of individuals both on the job and while conducting activities of daily living (ADLs). Indeed, the National Safety Council (2002) reported that slips and falls are the leading cause of death in the workplace, as well as account for more than 20% of all disabling injuries (Yoon & Lockhart, 2006). According to the U.S. Bureau of Labor Statistics (BLS), Slip, Trip and Fall (STF) events account for about 16% of all work-related accidental deaths (Bureau of Labor Statistics, 2016).

Apart from the personal toll that STF-related injuries incur, such incidents result in a significant economic toll in terms of lost wages and worker compensation claims. Liberty Mutual indicated that workplace injuries cost businesses approximately $1 billion/week. Moreover, the latest 2016 Liberty Mutual Workplace Safety Index indicated that total cost of disabling workplace injuries amounts to $61.88 billion per year, of which 28.9% ($17.92 billion) is associated with injuries due to STF (Liberty Mutual Group, 2016)**.** Moreover, the BLS revealed that every year approximately one million Americans experience an STF injury, and that employers spend in the range of $40,000 per STF-related incident (Department of Health and Human Services/NIOSH/BLS 2010). In fact, for most industry groups, slips and falls account for among the highest compensation claims by workers (Leamon & Murphy, 1995). The National Safety Council estimated that compensation and medical costs associated with employee slip and fall accidents total approximately $70 billion/year (National Safety Council, 2015). Also, the types of compensation claims from an evaluation study of hospital employees reveal that of the total 2,263 claims, overexertion or other bodily accommodations from trying to keep from hitting the ground (caused as a result of slip/trip without actually falling) accounted for highest number of claims

(34.46%) (Bell et al., 2008). As a proportion of total spending on workers' compensation claims (2005-2009), a full 27% were related to STF injuries (OSHA, 2013).

1.1.2. Gait Cycle

In order to better study the risk of STFs, it is imperative to have a comprehensive understanding of the human gait cycle. Gait is defined as a manner of walking or a sequence of particular movement of steps by which an individual moves forward (Lockhart, 2013). The human gait cycle is divided into two main phases: stance and swing (Figure 1). The stance phase accounts for 60% of the gait cycle, while the swing phase accounts for the remaining 40% (Bhattacharya & McGlothlin, 2012).

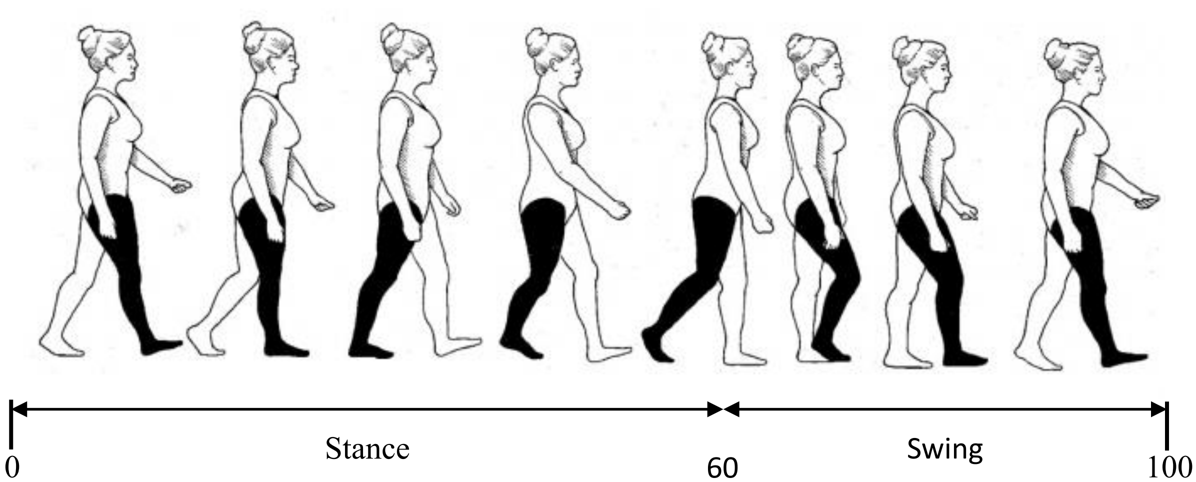

Figure 1: Human gait cycle during normal walking

Obtained from http://www.anatomy-physiotherapy.com/articles/systems/musculoskeletal/postural-dependence-of-locomotion-during-gait-initiation

While the gait cycle described below is for the right leg, it can also be generalized to the left. For the stance phase (during which the foot maintains contact with the ground), there are three sub-phases:

1) Contact/Heel Strike: Begins with the heel strike, during which the right leg takes the full weight of the body; this phase is also known as loading response, and concludes once the other foot (left foot) lifts off the surface.

2) Mid Stance: The right foot gradually begins to lift from rear and continues until the left foot makes initial contact with the ground, while preparing for the next phase.

3) Propulsion: This phase begins after the left foot makes contacts with the ground, followed by heel lift, and continuing through toe-off.

The swing phase refers to the period when the foot is off the ground. The principal task involved in the swing phase is to help the foot recover from toe-off, while preparing for the coming heel strike. Swing consists of the following two sub-phases:

1) Early Swing/Pre-swing: This starts at toe-off, with both feet in contact with the ground simultaneously.

2) Late Swing/Terminal Swing: In this sub-phase, the foot recovers from toe-off and sets itself into a stiff position in preparation for contact with the walking surface (Maynard & Curry, 2005; Bhattacharya & McGlothlin, 2012).

1.1.3. Initiation of Slip and its Types

Understanding the gait cycle is essential for interpreting the conditions under which a slip or fall could occur. Specifically, heel strike and toe-off represent two important phases of the human gait cycle that are more likely to precipitate a STF-related incident. Note that maximum friction occurs during the initial portion of the contact sub-phase. As the heel strikes the ground at an angle, two forces are imposed: one directed vertically downwards and the other directed forward in the direction of travel. Slip is initiated when the coefficient of friction between the shoe sole and the surface is lower than the ratio of the two force components i.e., horizontal and vertical force components (Maynard & Curry, 2005). Hence, having knowledge about gait cycle helps in classifying the types of potential STFs, which are discussed in the following sections.

Slips can be broadly classified in two categories: (a) a forward or backward slip, and (b) a microslip, slip, or slide (based on the length of the slip).

a) <u>Forward slip and backward slip:</u> A forward slip on the leading foot typically occurs during heel contact; in contrast, backward slip on the sole forepart typically occurs during the toe-off. A backward slip is considered less dangerous than a forward slip since the weight on the trailing foot is being quickly transferred to the heel of the leading foot during the toe-off phase. However, because a forward slip is initiated at back edge of the heel during the heel landing phase, it would be more likely to result in a dangerous fall because the entire weight of the body is transferred to the leading foot (Bakken, LaRue, Hyde, Abele, & Cohen, 2007).

b) <u>Microslip, Slip and Slide:</u> A "microslip" is defined as a slip that is shorter than 3 cm; a slip is generally between 8 and 10 cm; and a slide refers to the uncontrolled movement of the heel, which will likely occur as a result of a longer slip (i.e., more than 10 cm). Microslips are normally unreported, and a slip will lead to rapid corrective efforts made for regaining balance. A slide, however, is more likely to result in a fall because of the loss of balance (Chang, Leclercq, Lockhart & Haslam, 2016).

1.1.4. Initiation of a Trip and its Types

A trip can occur when the foot collides (strikes or hits) with an object, causing the person to lose balance. More particularly, a trip will occur when the lower leg or foot (the one which is in swing phase) hits an object lying on the ground while the upper part of the body continues to move forward, resulting in a loss of balance. One study has confirmed that walking surface irregularities as minimal as 5 mm can be sufficient for a person to trip (Begg, Best, Dell'Oro, & Taylor, 2007). A trip hazard exists when there is an exposed vertically oriented surface either above or below the primary ambulation surface plane that projects from it, but is not necessarily connected to it. This condition might lead the person to strike the surface with the foot and incur injury because of the trip event. Tripping falls are relatively rapid falls that can occur with or without the presence of a vertically oriented surface. Typically, a trip that occurs in the absence of a protruding vertically oriented surface occurs when the individual strikes his or her foot (or some other support base component) against the walking surface, resulting in a stagger and a fall. Generally, this event is referred to as a stumble (Bakken et al., 2007).

1.1.5. Initiation of Fall and its Types

Falls are common among all age groups and can occur in virtually any occupation, as well as at home and during leisure-time activities (Chang et al., 2016). Mostly, falls are initiated due to loss of balance resulting from a slip, trip, or stumble. Slipperiness/slipping accounts for 40-50% of all fall-related injuries (Courtney, Sorock et al.2001). Falls are broadly classified as follows:

Free fall: In this type of fall (occurring mainly from slipping), the victim's body completely loses contact with the walking surface prior to impact, resulting in an unimpeded downward acceleration of the body.

Rotational falls: In this type of fall (occurring generally due to tripping), the victim is incapable of moving his or her feet or legs forward to reposition the center of gravity within the support base after the incident, such that the upper body rotates about the support base and falls to the ground.

Crumple falls: This type of falls typically occur when the individual encounters a misstep hazard with low walking speed. A crumple fall occurs because the body's neuromusculoskeletal responses may not be sufficient for maintaining upright stability.

Tumble falls: This type of fall results from the body's failed attempts at fall prevention, combined with the person assuming (to the best of their ability) a "fall position" during the injury mitigation phase of the fall (Bakken et al., 2007).

Data collected by the Bureau of Labor Statistics states that of the 4,904,055 injuries reported by major US industries during the 1999-2001 period, 18.14% (889,816) were due to falls (same level fall, lower level fall, and unspecified falls) (Yoon & Lockhart, 2006).

1.2. Factors contributing to Slip, Trip and Fall

Given the range of potential short- and longer-term consequences of STF accidents in various industries, it is important to focus on factors that lead to such accidents. Considering a typical industrial workplace, there are several factors that may engender STF incidents. Broadly, they typically comprise one or more of the following factors (Bentley & Haslam, 2001):

Individual factors: age, sex, training issues, awareness of safety issues, etc.

Equipment and processing factors: pace of walking, shoe material, walkway design, type of flooring, etc.

Environmental factors: weather, slippery/wet floor, proper lighting, sudden highs and lows during walking, etc.

The Center for Disease Control and Prevention (CDC) has comprehensively categorized the potential hazards as follows: (a) floor contamination (food, water, oil, grease), (b) substandard drainage facilities, (c) anomalies on the walking surface (both inside and outside the workplace), (d) climate conditions (snow & ice), (e) insufficient lighting condition, (f) handrails and stairs, (g) tripping hazards (unidentified objects lying around, tangled cords and cables), and (h) inappropriate use of floor mats (Centers for Disease, 2011).

Further intensifying the potential danger associated these above-mentioned factors is human fatigue, which represents another significant contributor to increasing the chances of a slip-related fall. The literature typically classifies fatigue into two types: cognitive/mental and muscle fatigue.

<u>Cognitive/Mental Fatigue:</u> Mental fatigue can occur when a person performs highly repetitive tasks for a prolonged duration. Lew and Qu (2014) observed the adverse effects of people with mental fatigue compared to people with no fatigue in terms of their ability to perform work. The researchers confirmed that mental fatigue increases the possibility of slip initiation, poorer slip detection, and decreased reactive responses while slipping. Hence, cognitive fatigue is a significant factor that increases the likelihood of slips and falls.

<u>Muscle Fatigue:</u> Localized muscle fatigue (LMF) results from any repetitive task involving the use of some specific muscle/set of muscles. Kinetic and kinematic data has shown that LMF increases the risk of slip-induced falls (Parijat & Lockhart, 2008; Lew & Qu, 2014). Moreover, LMF also causes a delay in the reactive response required to recover from a fall. Hence, muscle fatigue is classified as one of the significant factors responsible for initiation of STF.

1.3. Slips and Trips without fall

As indicated in the prior sections, there is a significant body of research pertaining to the risk of falls due to slips, trips, and stumbles. What remains underreported are the bodily risks associated with slips/trips that do not result in a fall (STWF), but rather cause low-back injuries due to the required effort to regain one's balance from slip or trip. Some statistics do exist that provide information about the significance of STWF and associated compensation costs. Based on an analysis conducted by The European Commission (2008), there were 3,983,881 non-fatal accidents reported at the workplace during 2005, involving at least 3 days of absence from work. Of these accidents, "slipping - stumbling and falling - fall of a person on same level" was the largest reported category, constituting 14.4% (573,679). Further, 4.4% (175,291) were reported as "treading badly, twisting leg or ankle, slipping without falling." Based on a similar study conducted by the BLS (2014), there were 1,162,210 non-fatal occupational accidents and diseases reported in 2013 at private companies and government agencies - 17.4% (202,225) of which were falls on the same level resulting in a median loss of 10 work days. Further, 4.4% (51,138) of reported injuries were slips or trips without a fall, but leading to low-back injury, resulting in a median loss of 11 work days (Chang et al., 2016).

Amandus, Bell, Tiesman, and Biddle (2012) conducted a four-year study (Jan 2004 through Feb 2008) involving 4,070 workers in a helicopter manufacturing plant, of which 2,378 were reportedly injured in one way or another. Among these 2,378 injuries, a total of 226 STF-related accidents were reported, of which 46 were falls (20%) to a lower elevation level, (e.g., from stands or large machinery), 117 (52%) were falls on the same level, 41 (18%) occurred from loss of balance without a fall, and 22 (10%) from other events. The helicopter manufacturing plant had incurred a heavy compensation cost of $1,543,946 due to injuries. Data collected by BLS indicates that of the 1,537,567 injuries and illness reported in major private industries in the US (2001), 50,269 (3.3%) resulted from slips in the workplace. Additionally, of these over a million-and-a-half reported injuries and illness, 42,679 (2.8%) had injured their back (Yoon & Lockhart, 2006). It must be noted that data for occupational injuries are available only for a limited few countries.

Nonetheless, because of the nature of occupational injuries (for example, STWF injuries often go unnoticed or unreported), it is entirely plausible that workers across the globe are at the same risk.

1.4. Low Back Injuries due to Slips and Trips

Slips and trips that do not lead to falling, also called “near-accidents,” are known to be hazardous to the spine - potentially due to the rapid corrective movements made to restore balance. Such movements can possibly initiate substantial muscle forces, as well as harmful loading on the spine (Lavender, Sommerich, Sudhakar, & Marris, 1988). For example, it has been observed that lower-extremity joint moments increased significantly during slipping compared to those during normal walking (Cham & Redfern, 2001). Researchers have reported that of the various underfoot accidents, slipping accounted for 62%, tripping for 17%, and ankle twisting for 12% (Manning, Ayers, Jones, Bruce & Cohen, 1988). Further, 12% of these accidents led to lumbosacral injuries.

Paradoxically, although the overall number of occupational injuries has been declining in industrially developed countries, injuries due to slips have increased. For example, Chang et al. (2016) examined work-related accidents that resulted in lost work days in French companies operating within the country’s general social security system over the period 1987-2011. While the authors noted an overall reduction of 13.6 accidents/1000 employees during this period, the reduction in number of injuries due to slips and trips (excluding falls from height) was a meager 1 accident/1000 employees. Depending upon the severity of a slip, contusions and crushing can also occur - possibly in combination with low-back injuries. Bentley and Haslam's (2001) study of postal delivery workers who experienced some form of injury in the performance of their job most frequently reported ankle injuries (23%), followed by knee (17%) and back (16%). Indeed, ankle and back problems lead to almost 50% of lost workdays - with the former resulting from trips, and the latter resulting from falls (Chang et al., 2016).

Considerable compensation costs are associated with low back injuries due to STF. Murphy and Courtney (2000) reported that 11% of low back pain-related claims can be attributed to slips and falls. In a comprehensive study on STF covering the period 1996-2005, of the total 472 compensation claims by workers for STF-related injury, 185 (44.9%) claims involved lower extremity injuries (knees, ankles, feet) and 73 (16.2%) were involved injuries to the back or trunk (Bell et al., 2008). Guo, Tanaka, Halperin, and Cameron (1999) reported that in the industrial environment, back injuries represent the most frequently cited cause of worker compensation claims in the United States.

Moreover, OSHA's European study report states that, "between 60-90% of all people will suffer from Low Back Disorders (LBD) at some point in their lives" (European Agency for Safety and Health at Work, 2000). The prevalence of low back pain has been reported to be "55-87% throughout one's lifetime" (Videman, Nurimen, Tola, Kuorinka, Vanharanta, & Troup, 1984). Similarly, a recent study confirms that unanticipated and unexpected perturbations during human bipedal locomotion can be quite hazardous for the lumbar spine and could potentially lead to low-back pain development - potentially because of the rapid corrective movements required for recovery after a slip to regain balance (Liu, Lockhart, & Kim, 2014).

1.5. Research Gap

As noted earlier, the linkage between slips and trips and low-back disorders is well established (e.g., Murray, Mollinger, Gardner, & Sepic, 1984; Rowe & White 1996). To our knowledge, however, there is a lack of scholarly evidence regarding the loads acting on the low back due to incidences of slips and trips that do not result in a fall (STWF). It is, of course, one's natural instinct to attempt to keep from hitting the ground (if at all possible) after a slip or trip - but such efforts are known to initiate substantial muscle forces, as well as harmful loading on the spine, due to those split-second corrective actions (Lavender et al., 1988). Given the dearth of quantifiable evidence as to the level of low-back loading during dynamic events like STWF, additional evidence is needed to elucidate this relationship. Thus, the goal of the current study is to quantify the lumbar kinematics, lumbar muscle activity and lumbosacral reaction forces and moments on the low back due to rapid corrective actions taken to prevent falling.

As a prior step in reaching this goal, we refer to a published conference paper in which preliminary results were presented with respect to lumbar muscle activity, kinematics, and kinetics of the low back due to induced slips (Rashedi, Jia, Nussbaum, & Lockhart, 2012). In the current study, we are expanding that effort in an effort to overcome some of the limitations of the prior investigation. Specifically, only six individuals participated in the earlier study, which affected the power of statistical analysis and potentially the generalizability of our findings. Moreover, in addition to an induced slip as a perturbation, here we include trips as another important source of human gait perturbation that might result in substantial corrective balance efforts. Importantly, this investigation was also designed to improve the validation process for the biomechanical model for highly dynamic conditions during gait perturbation. Previously, we validated the model outcomes by comparing the low-back moments obtained from the model to the similar findings in the literature during normal walking (i.e., less dynamic conditions). In the current study, we will validate the predicted moments during perturbed gait trials using the “Inverse Dynamics” methodology, which is briefly described in the next section.

1.6. Innovation

As detailed earlier, prior studies have evaluated muscle activity and lumbar kinematics during normal human gait (e.g., Murray et al., 1984, Rowe & White 1996). In contrast, there is far less information in the literature investigating how loads act on the low back due to more dynamic activities like STWF. Along with analyzing the effects of a slip on the low back, the current study assessed the effects of another important perturbation to human gait: a trip. Again, to the best of our knowledge, no prior study has assessed the lumbar kinematics, lumbosacral loads and related muscle activity due to unexpected trips during walking. Accordingly, the current study utilized an EMG-based lumbar spine model to obtain lumbar kinematics, lumbar muscle activity and low-back loads. Another innovative aspect of this study is incorporating the use of inverse dynamics analysis to validate the obtained lumbosacral reaction moment from the EMG-based model. Inverse dynamics refers to the backward calculation of loads compared to forward dynamics, which determines the displacement of an object by integrating the known loads. Conversely, inverse dynamics calculates the loads acting on an object by differentiating the known displacement. Inverse dynamics determines the desired net joint loads by computing the kinematics (motion) and kinetics (forces that cause motion) using Newton's laws of motion (Robertson, Caldwell, Hamill, Kamen, & Whittlesey, 2014).

1.7. Hypothesis

To reiterate, unexpected surface perturbations that cause a walker to slip or trip - but not fall - could potentially produce a considerable impact on the kinematics, muscle activity and lumbosacral loads to that individual's low back. In this study, we seek to evaluate and quantify said kinematics, muscle activity and lumbosacral forces and moments due to STWF. Low-back loads calculated during the slip-and-trip trials were then compared with analogous results obtained during normal walking. Following hypotheses guided this investigation:

- Lumbosacral reaction loads (L5/S1), lumbar kinematics, and lumbar muscle activity would increase significantly during slip events compared to normal walking.

- Lumbosacral reaction loads (L5/S1), lumbar kinematics, and lumbar muscle activity would increase significantly during trip events compared to normal walking.

2. Methodology

2.1. Participants

A total of twelve participants (six males and six females) with a mean (SD) age of 23.67 (2.74) years, a mean height of 170.71 (7.67) cm, and a mean body mass of 63.32 (9.25) kg, were recruited for the experimental study conducted in the locomotion lab at the Grado Department of Industrial and Systems Engineering at Virginia Tech. Moderately physically active participants (exercising at least two times per week) with no history of neurological problems or recent lower extremity and low back musculoskeletal injury were recruited to avoid potential biasing of the study's findings. Prior to conducting the experiment, a general overview regarding the background and the goals of the study was described to participants, after which they were given a detailed explanation of their role and what they would be required to do. Participants were then asked to read and sign the informed consent form approved by Virginia Tech's Institutional Review Board (IRB).

2.2. Experimental Setup and Data Collection

Before beginning the actual experimental portion of the study, appropriate muscle sites were identified for electrode placement. Then, electrodes were attached on the dorsal and frontal part of each participant's trunk for measuring muscle activity via electromyography (EMG). First, however, the skin sites (a muscle-electrode interface) were fully prepared by rubbing gently with sandpaper and then cleaning the site with alcohol wipes to remove the dead skin cells. This process increased the likelihood for obtaining a high-quality signal and minimizing noise interference in the signal. Muscle activity was determined using EMG data recorded from 14 bilateral flexor and extensor muscles in the dorsal (Figure 2a) and the frontal (Figure 2b) region. The muscle activity was recorded at a frequency of 1000 Hz, band-pass filtered at 10-400 Hz (1^{st} order, butterworth), then rectified and low pass filtered at 2 Hz. The following muscles were used for measuring EMG activity: multifidus (MF), longissimus thoracis pars lumborum (LTL), iliocostalis lumborum pars

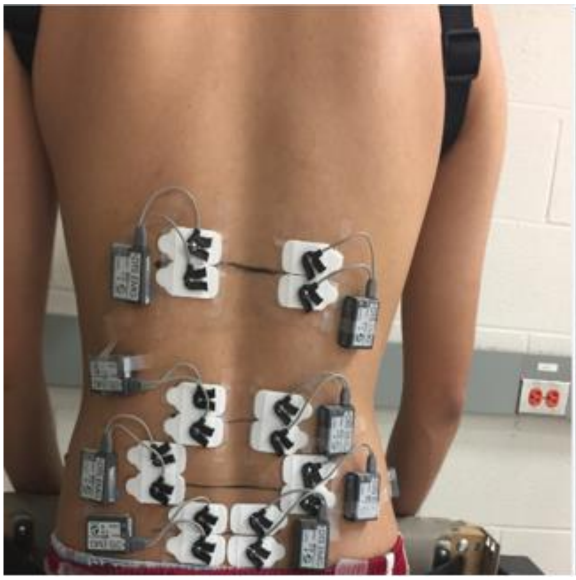

Figure 2a: Placement of electrodes (dorsal)

Figure 2b: Placement of electrodes (frontal)

Figure 3: Measurement of muscle activity during rest

lumborum (ILL), longissimus thoracis pars thoracis (LTT), external oblique (EO), internal oblique (IO), and rectus abdominis (RA) (Jia, Kim, & Nussbaum, 2011).

The first phase of the experiment involved assessing the relationship of MVC (Maximum Voluntary Contraction) and low-back moment in different directions. Subjects were first asked to lie down on a floor mattress in both supine and prone positions to measure muscle activity during rest. Figure 3 represents EMG being measured for one of the participants at rest in a prone position. After recording the muscle activity during rest, participants were asked to stand on a customized setup fixture that consisted of a force plate (AMTI ORG-7-1000, Watertown, MA, USA) attached to the base, as shown in Figures 4a/b.

Figure 4a: Measurement of MVC during axial rotation

Figure 4b: Measurement of MVC during right lateral bending

Three sets of maximum exertions were measured, during which subjects were required to reach their maximum level of muscle activity gradually in about 4-5 seconds without any jerky exertions. MVCs were measured from maximal voluntary muscle activation that involved trunk flexion/extension, axial rotation (both clockwise and counterclockwise) and left/right lateral bending. It should be noted that the participant’s movements were restricted by attaching straps to

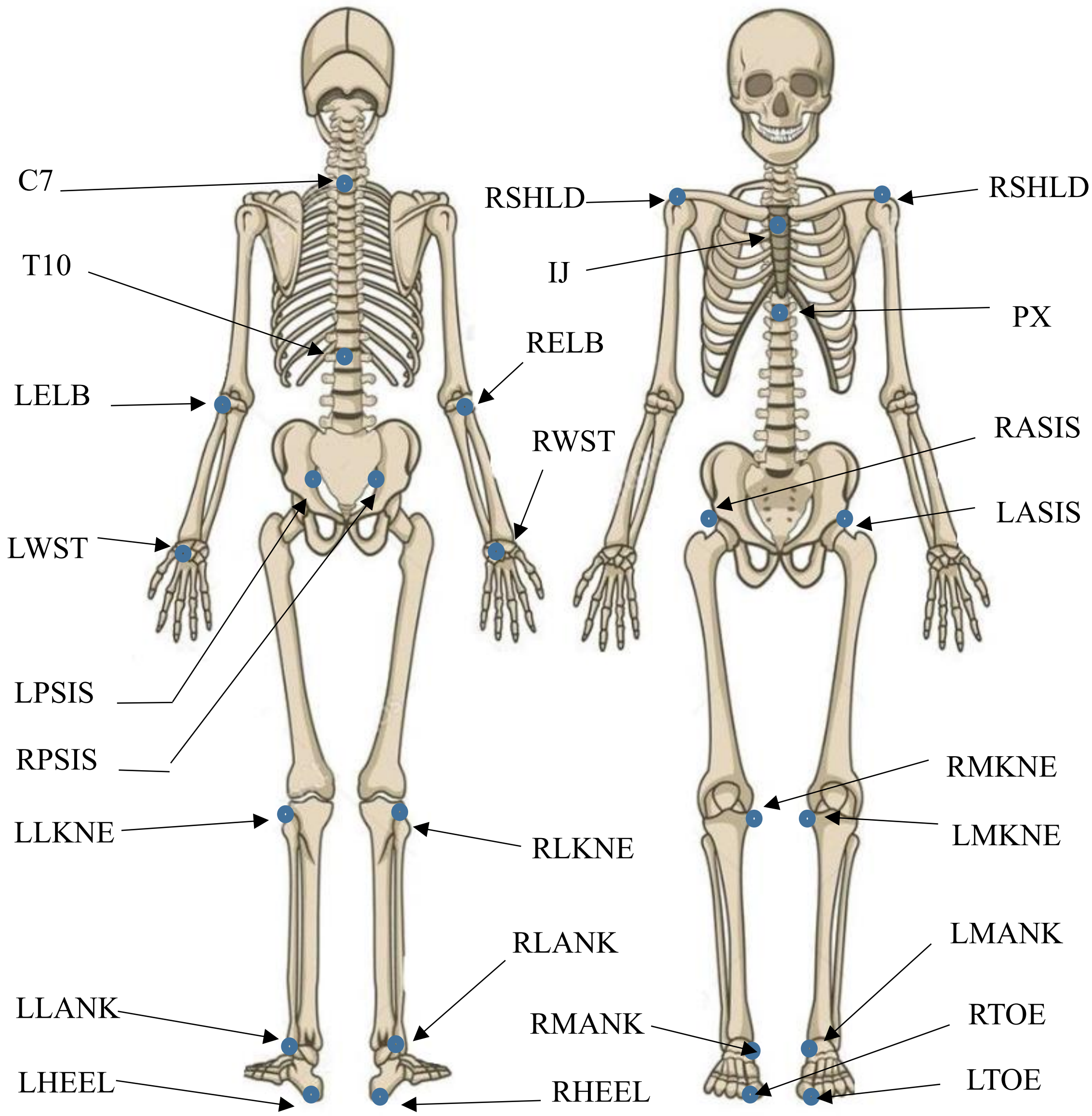

Figure 5: Placement of markers on the bony landmarks

Obtained from http://ru.depositphotos.com/2866723/stock-illustration-human-skeleton-in-vector.html the shoulder, pelvis and knee. Figure 4a and Figure 4b depicts the measurement of MVC during axial rotation and right lateral bending, respectively. Except for the extension trials, which were performed with around 20-degree trunk flexion, all other exertion trials were performed in an upright posture (Rashedi et al., 2012).

During the second phase of the experiment, 26 passive reflective markers were placed on each participant's bony landmarks, including on the foot, ankle, knee, pelvis, trunk, wrist, elbow and shoulder (Davis, Ounpuu, Tyburski, & Gage, 1991; Damsgaard, Rasmussen, Torholm, Christensen, Surma, & de Zee, 2006). Figure 5 provides a schematic diagram of the 26 reflective markers placed on the bony landmarks of the participant and their labelling. A three-dimensional, seven-camera motion capture system (Vicon Mx, Vicon Motion Systems Inc, Denver, Co, USA) was used to record the marker trajectories. The marker data was recorded at a frequency of 100 Hz. The Vicon motion capture system was oriented in a GCS (global coordinate system) with +X-Axis towards the direction of the walking, +Z-Axis facing upwards (towards the ceiling), and +Y-Axis towards the left using the right-hand thumb rule.

Participants were then asked to walk on the customized walkway designed for each of the three experimental trials (unperturbed, slip, and trip). The walkway for the participants incorporated two force plates (AMTI ORG-7-1000, Watertown, MA, USA) as a part of the experimental setup. For each participant, several walking trials (without any perturbation) were recorded, after which that data was compared with data obtained from the unexpected slip and trip trials. A sliding platform partly covered with lubricant and a trip-inducing mechanism consisting of a trip-plate was customized and built on the walkway as shown in Figure 6.

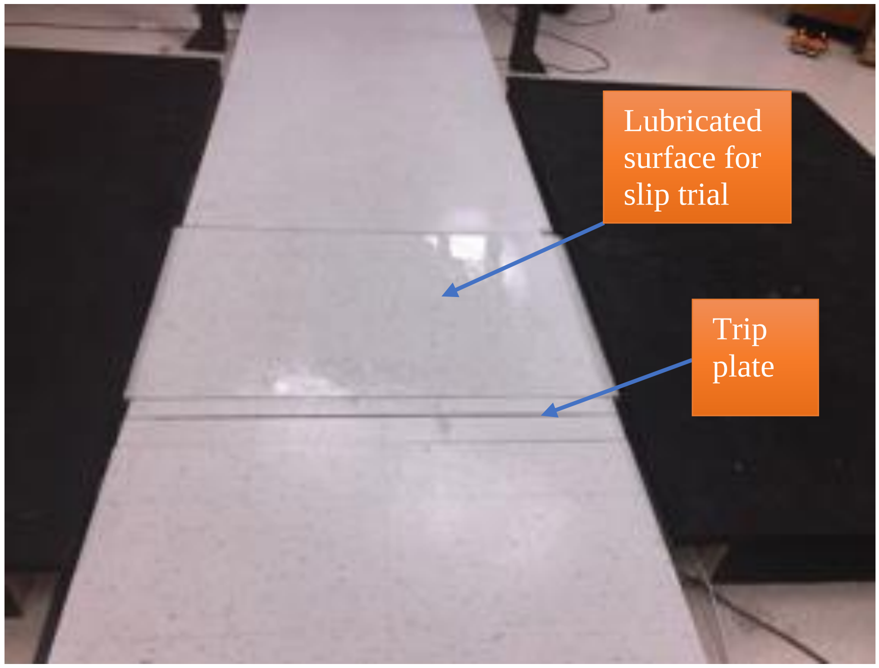

Figure 6: Slippery surface with sliding mechanism and trip plate

Figure 7: Sliding mechanism for the slippery surface

For the slip trial, a platform with a sliding mechanism integrated with the walkway (Figure 7) was activated by pulling a cord. The sliding platform was partly covered with slippery liquids. While executing the slip trial, the cord was pulled in order to bring the slippery part of the sliding surface right at the center of the walkway. Figure 8 shows a participant experiencing a slip and trying to recover while walking on the slippery surface. Note that the individual is prevented from falling through use of a harness.

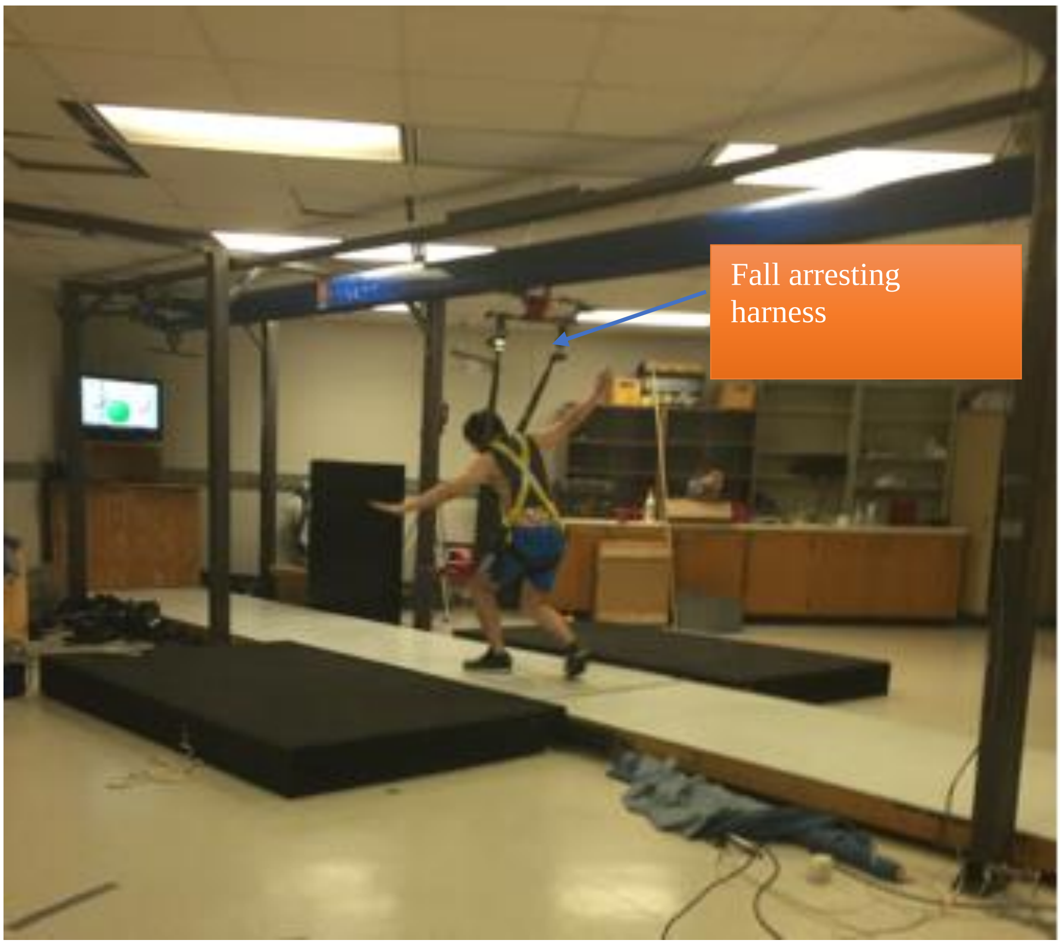

Figure 8: Participant experiencing a slip

Figure 9: Pulley mechanism used to activate the trip plate

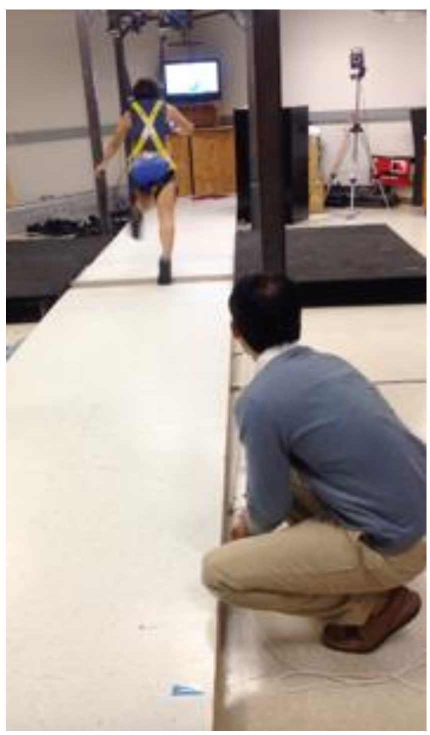

Figure 10: Participant experiencing a trip due to activating the trip plate by pulling a cord

The trip-plate consisted of a pulley mechanism (Figure 9) attached to the trip plate (4 feet in width and 5 cm in height), which could be activated by pulling a cord that positioned the trip-plate in an upright position. Before the trip trial, each participant's gait cycle was carefully examined and adjusted by integrating several unperturbed walking trials. This measure ensured that the participant's foot landed precisely just before the trip-plate, thereby inducing a successful trip. Figure 10 demonstrates one of the participants experiencing a trip after her right foot encounters the trip plate activated by pulling a cord. For ensuring participant safety, the walkway was equipped with an overhead harness system for arresting falls in case the individual could not recover from the induced slip or trip (Cham & Redfern 2001; Lockhart et al., 2005). To standardize the experimental process, all the participants wore the same type of shoe (fitted properly for each individual) during all trials.

Important for this investigation was the need to introduce a level of unexpectedness in order to simulate an actual STWF that might occur in a real-life situation at work or at home. To do so, we incorporated a number of distraction activities between the trials (unperturbed, slip and trip trials) to divert attention from an upcoming slip/trip perturbations.

- A simple counting task was designed with a monitor screen mounted at the far end of the walkway that flashed circles of different colors (red, bl]ue and green) in random order. First, participants donned headphones that played music - and while walking, they were required to count the number of times a specific color circle flashed until they reached the far end of the walkway near the monitor. They were asked to wait there for further instructions before returning or removing the headphones (Cham & Redfern, 2001).
- At the other end of the walkway (the end opposite to the monitor screen) there was an additional "distracting task" that was integrated intermittently between all the trials (slip, trip, and unperturbed). A stack of four different-colored letter papers were kept side-by-

side on the desk and participants had to make a pile by collecting one letter paper of each color as quickly as they could.

Again, the purpose of these tasks was to keep participants occupied in some activity to provide sufficient time for experimenters to incorporate unforeseen changes to the walkway that would simulate as closely as possible an actual unexpected STWF while walking. Once the slip-and-trip trials were completed, participants were asked to report the slip and trip expectancy rating on a scale from zero to ten - with zero indicating that the participant had experienced a totally unexpected slip or trip while walking. The average slip and trip expectancy rating across all participants was reported as 0.9.

2.3. Biomechanical Modelling and Analysis

An important aspect of this investigation involved the development of a biomechanical model to analyze the obtained experimental data. Researchers have developed a number of models that feature invasive methods as a means to estimate *in vivo* muscle forces (Rohlmann, Arntz, Graichen, & Bergmann, 2001). Conversely, a variety of non-invasive biomechanical models have been developed that target multiple muscles (to account for the muscle cocontraction phenomenon) for measuring spinal loads. Multi-muscle biomechanical models have employed diverse strategies to distribute loads over several muscles. These include optimization (van Dieën, 1997; Cromwell, Schultz, Beck, & Warwick, 1989); electromyography (EMG) (Granata & Marras, 1995; Khoo, Goh, & Bose, 1995; Nussbaum & Chaffin, 1998); stochastic models (Mirka & Marras, 1993); neural networks (Nussbaum & Chaffin, 1996); and EMG-optimization hybrids (Cholewicki & McGill, 1994). Of these different types of models, EMG-based models have been used most frequently because of their inherent proclivity to include different levels of muscle cocontraction.

The flow chart constructed in Figure 11 represents the biomechanical model, its inputs, and outputs. The main inputs to the biomechanical model included EMG activity and motion data, as detailed below.

<u>EMG Activity:</u> Each participant's muscle activity was obtained using fourteen surface EMG electrodes placed on the lumbar and the belly muscles. The recorded EMG data was down-sampled from 1000 Hz to 100 Hz to match recorded marker data and synchronize the data-analysis process. Muscle activity was recorded for all the trials including normal walking, slips, and trips.

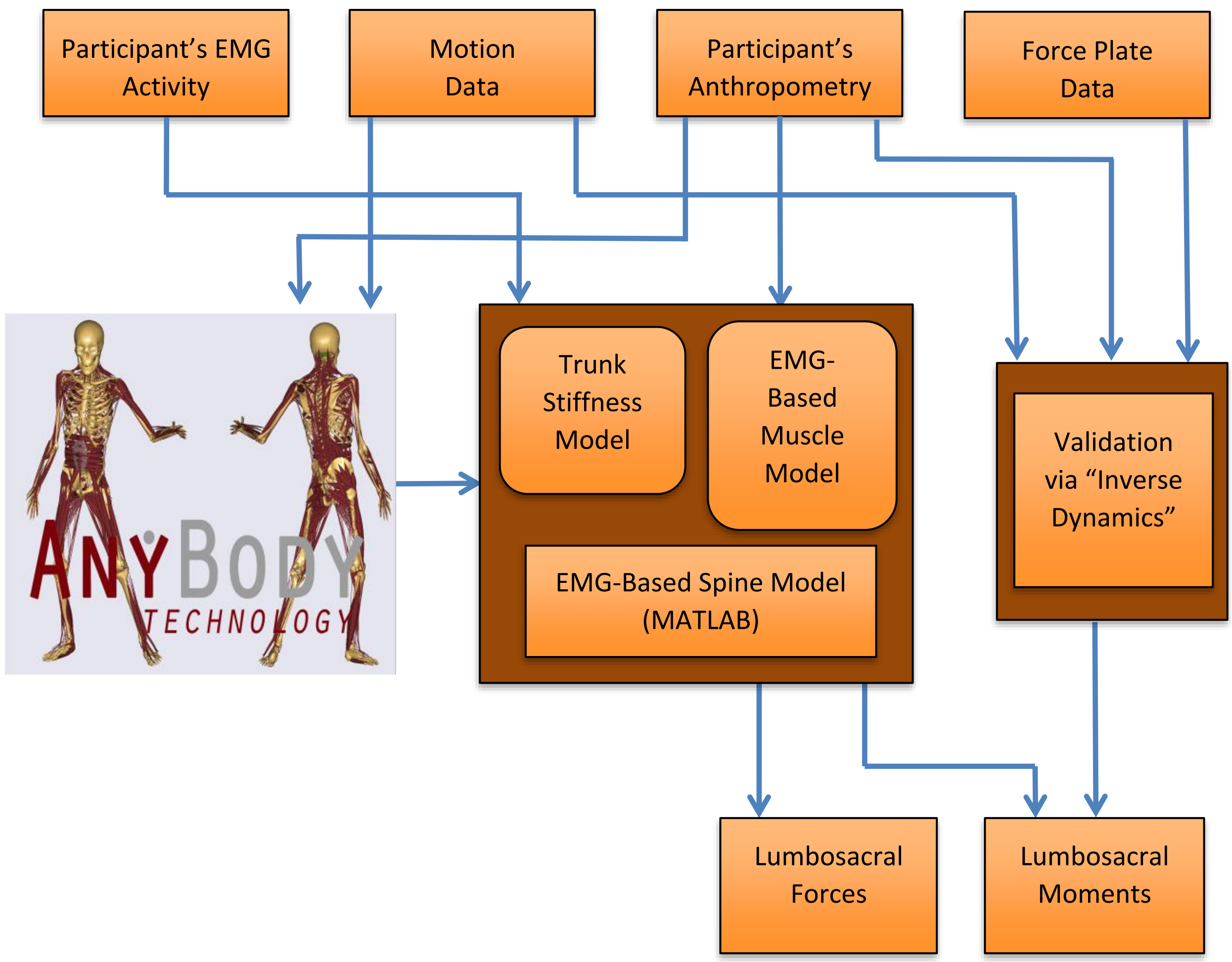

Figure 11: Flow chart for biomechanical analysis

Motion Data: A MATLAB model was developed using motion data recorded from the marker trajectory. The marker data recorded in the GCS was converted to the coordinate system defined by the anatomical model developed in AnyBody musculoskeletal modeling system (v5.0, AnyBody Technology, Aalborg, Denmark). AnyBody's coordinate system was oriented with +X-Axis towards the direction of the walking, +Y-Axis facing upwards (towards the ceiling) and +Z-Axis towards the right using the right-hand thumb rule. The marker data was converted from GCS to ACS (AnyBody coordinate system) using the following conversion:

$$ACS\_X = GCS\_X \quad (1.1)$$

$$ACS\_Y = GCS\_Z \quad (1.2)$$

$$ACS\_Z = -GCS\_Y \quad (1.3)$$

An embedded coordinate system was determined for a rigid body segment with at least three non-collinear markers. The pelvis coordinate system was defined using the three-dimensional location vectors of the four pelvic markers: RASIS, RPSIS, LASIS & LPSIS. Vectors A_1 and A_2 were defined as:

$$A_1 = 0.5\times(LASIS + RASIS) - 0.5\times(RPSIS + LPSIS) \quad (1.4)$$

$$A_2 = (RASIS - LASIS) \quad (1.5)$$

The vector A_2 was normalized to obtain unit vector U_1. The vector A_3 was then defined using Gram-Schmidt orthogonalization procedure (Davis et al., 1991):

$$A_3 = A_1 - (A_1.U_1)\times U_1 \quad (1.6)$$

The vector A_3 was further normalized to become unit vector U_2. Transformation matrix, which defines the orientation of pelvis relative to the GCS, was developed to determine Euler angles with an Y-X-Z rotation (Davis et al., 1991). These angles relate to flexion/extension, abduction/adduction, and internal /external rotation, respectively, and were computed using the following relationship:

$$\theta_x = -sin^{-1}\left[U_1.GCS_y\right] \quad (1.7)$$

$$\theta_y = sin^{-1}\left[\frac{U_1.GCS_x}{cos\,\theta_x}\right] \quad (1.8)$$

$$\theta_z = sin^{-1}\left[\frac{U_2.GCS_y}{cos\,\theta_x}\right] \quad (1.9)$$

Similarly, the joint rotation angles were obtained for the trunk segment using the IJ, C7 and PX markers. The pelvis angles were calculated as absolute angles (referenced to the GCS), whereas the trunk angles were calculated relative to data obtained for the pelvis.

Anthropometric Data: Before the start of the experimental trials, anthropometric data was recorded for all participants with respect to body mass, height, chest width, chest depth, neck height, L5/S1 height, shoe height, shoe size, and dominant foot (left or right).

Force Plate Data: Force and moment data were recorded for all participants utilizing the two force plates incorporated into the walkway designed for the experimental trials.

Force plate data, lumbar kinematics, and anthropometric data were used as inputs to the anatomical model developed for the AnyBody musculoskeletal modeling system. The AnyBody repository with predefined values was used to extract initial insertions, via points, and the origin of a total of 92 muscle fascicles (values were scaled based on each participant's anthropometry). Using the marker data, the AnyBody model calculated lumbar kinematics and the lengths, moment arms, and velocities of the muscle fascicles. A 3D, dynamic, EMG-based MATLAB model of the lumbar spine used the output from AnyBody model, along with normalized EMG and participants individual anthropometry (Jia et al., 2011). As a result, output from the EMG-based MATLAB model provided the spinal loads, which included the lumbosacral (L5/S1) reaction forces and moments.

For validation purpose; the L5/S1 reaction moments from the EMG-based MATLAB spine model was compared to a 3D biomechanical model developed for the bottom-up inverse dynamics analysis (Erdemir, McLean, Herzog, & van den Bogert 2007; Robert, Chèze, Dumas, & Verriest, 2007; St-Onge, Côté, Preuss, Patenaude, & Fung, 2011; Shourijeh, Smale, Potvin, & Benoit 2016). A 3D inverse dynamics model was developed in Visual3D (C-Motion Inc., Germantown, MD, USA) consisting of seven segments defined by the markers: right and left feet, shanks, and thighs, as well as the pelvis. The pelvis was considered as a single rigid segment, defined by the four markers: RASIS, LASIS, RPSIS, and LPSIS. The posterior part of the pelvis was considered as the L5/S1 segment. Due to the absence of markers on the hip joint during data collection, the hip joint was defined by Visual3D based on the position of the pelvis segment (hip was considered proximal to the thigh segment). An inverse dynamics model was developed using the following information (Winter, 2009):

- Participant's body measurements obtained from anthropometric data.
- The center of pressure and ground reaction forces from the force plate data.
- Movement kinematics involving the marker data for determining the position of each body segment. These segments comprised the left and right feet, shanks, thighs, as well as a small section of the trunk to reach to the L5/S1 lumbar region level; note that for this investigation we assumed that each person's body segments were connected via frictionless spherical joints.

2.4. Statistical Analysis

The independent measures for this study included age, weight, height, gender, and condition (normal walking, slip, trip), while the dependent measures are categorized into lumbar kinematics (the motion of the trunk relative to the pelvis), normalized muscle activity (NEMG) and lumbosacral forces and moments. Furthermore, for our lumbar kinematics findings, the motion of the trunk relative to the pelvis was analyzed in three different directions (axial rotation, lateral bending and flexion-extension) along with the resultant lumbar motion. NEMG-related findings analyzed all fourteen individual muscles along with the mean NEMG. Similarly, the lumbosacral forces (A/P shear, lateral shear, and compression) and moments (lateral bending, flexion-extension, and axial rotation) were individually analyzed along with the resultant force and the resultant moment.

One-way repeated measure analyses of variance (ANOVAs) were conducted to assess the influence of gender, age, height, weight, walking condition (normal walking, slip, and trip) and their interaction on the response variables (muscle activity and lumbosacral loads). Two-factor interaction effects for gender and condition were included in the analyses to investigate the combined effect of the two components on muscle activity and lumbosacral loads. All the statistical analyses were conducted using JMP Pro 13.0 (SAS Institute Inc., Cary, NC, USA) with a significance level of 0.05. Participant's ID was nested with gender since each ID corresponded to one gender, and was assigned a random effect attribute. Since no significant effects were observed on the response variables due to age, weight, and height, these variables were removed from the model. As a result, the dependent measures of the model included gender, condition, and their interaction effect. Based on the results from statistical analyses, post-hoc comparisons were performed using Tukey's Honest Significant Difference (HSD) for assessing the differences between levels of statistically significant factors.

3. Results

A summary of the lumbar kinematics (the motion of the trunk relative to the pelvis), normalized muscle activity (NEMG) and lumbosacral forces and moments is presented in the following section. For the purpose of conciseness, representative outcomes are provided for one of the twelve participants, followed by summary of statistical analysis for all the participants.

3.1. Lumbar Kinematics

Three angles of rotation were used to represent the motion of the trunk relative to the pelvis (TRP) for normal walking, slip, and trip. These angles relate to axial rotation, lateral bending, and flexion-extension, respectively. As shown in Figure 12, the range of motion of the TRP was consistent before and after the right-foot heel contact during normal walking. For normal walking,

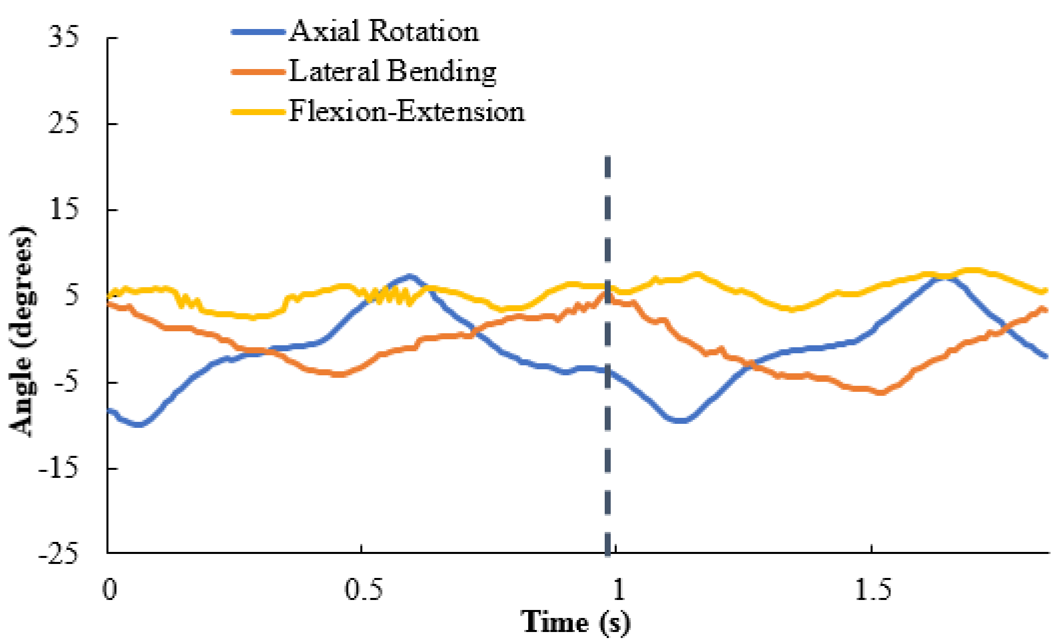

Figure 12: Motion of the trunk relative to the pelvis (axial rotation, lateral bending, flexion-extension) during normal walking. Dotted line indicates the heel contact on the force plate at 0.98 seconds.

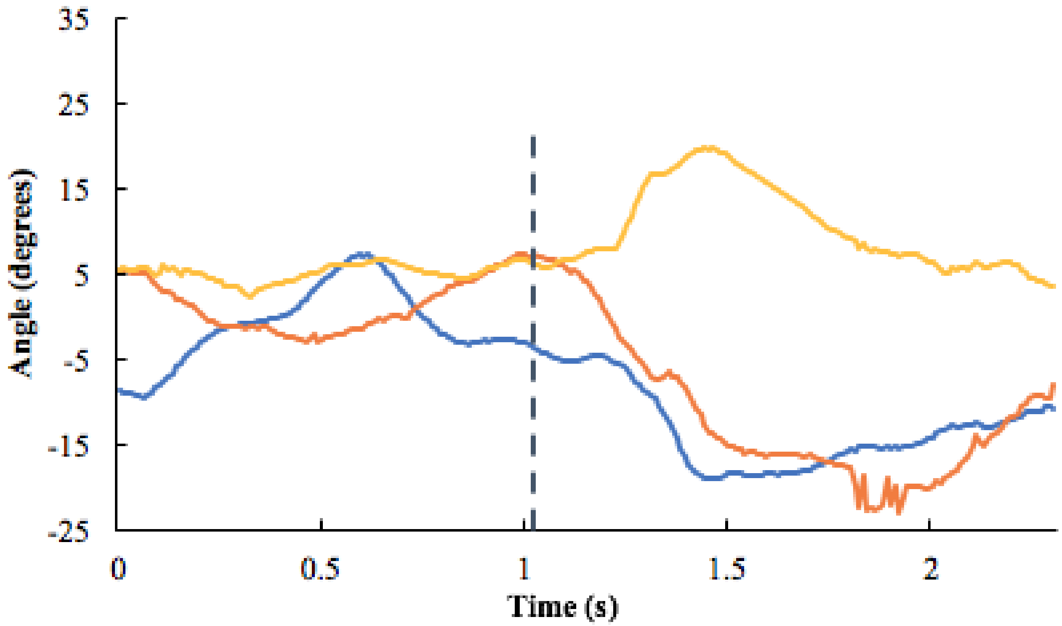

Figure 13: Motion of the trunk relative to the pelvis (axial rotation, lateral bending flexion-extension) during the slip trial. Dotted line indicates the heel contact on the force plate at 1.02 seconds.

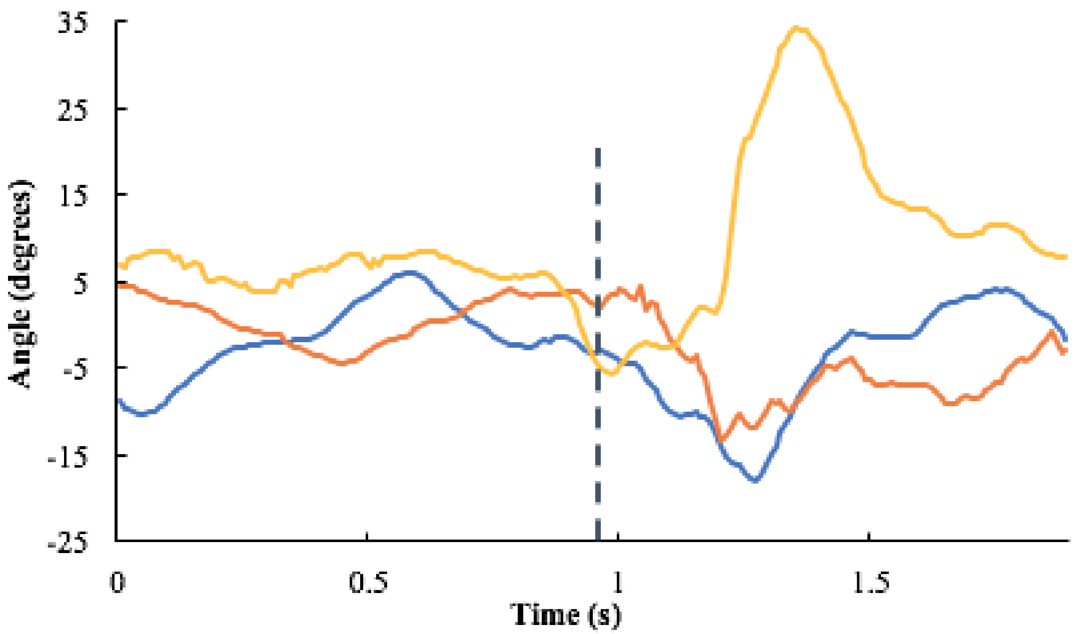

Figure 14: Motion of the trunk relative to the pelvis (axial rotation, lateral bending flexion-extension) during the trip trial. Dotted line indicates the heel contact on the force plate at 0.95 second.

all three rotation angles of the trunk relative to the pelvis followed a constant profile with a small overall variation.

In contrast, the flexion-extension angle of the trunk-pelvis was substantially increased during the slip trial (after the heel strike at 1.02 seconds, as shown in Figure 13). The range of flexion-extension angle for the slip trial was 17 degrees versus 6 degrees for normal walking. Additionally, there was a large increase in the lateral bending movement after the heel strike during the slip trial. The range of lateral bending angle for the slip trial was 29 degrees versus 11 degrees during normal walking. Likewise, after the heel strike, the axial rotation movement increased and the range of axial rotation angle for the slip trial was 25 degrees versus 15 degrees during normal walking.

Similarly, for the trip trial, the extension angle increased substantially after the heel strike at 0.95 seconds (Figure 14). The range of flexion-extension angle for slip trial was 37 degrees versus 6 degrees for normal walking. As seen in the slip trial, there was a large increase in the lateral bending movement after the heel strike during the trip trial. The range of lateral bending angle for the trip trial was 17 degrees versus 11 degrees during normal walking. The axial rotation movement increased after the heel strike and the range of axial rotation angle for the trip trial was 23 degrees versus 15 degrees during normal walking (Figure 12).

A similar range of angles and movement of the TRP was observed during normal walking, the slip, and the trip trials for the remainder of the participants. Table 1 provides a detailed summary regarding the mean (SD) of the range of the angle of TRP for all participants during normal walking, slip, and trip trials. Across all participants, the mean of the range of the resultant kinematics for the slip and trip trials showed more than a twofold increase compared to the normal walking. Table 2 provides a summary of the results from repeated measure ANOVAs of the angle

Table 1: Mean (SD) of the range of the angle of TRP (degree) during normal walking, slip, and trip trials across the different participants.

Angle of TRP	Normal Walking Mean (SD)	Slip Trial Mean (SD)	Trip Trial Mean (SD)
Axial Rotation	13.39 (2.69)	16.99 (6.55)	22.04 (4.69)
Lateral Bending	11.96 (3.44)	24.49 (9.21)	29.81 (10.87)
Flexion-Extension	5.57 (1.44)	32.54 (14.4)	24.71 (7.4)
Resultant Kinematics	19 (2.95)	45 (9.2)	42 (10.11)

Table 2: Summary of ANOVA results for the main and first-order interaction effect of gender and condition on angle of TRP.

Angle of TRP	Gender	Condition	Slip / Trip	Gender*Condition Interaction
Axial Rotation	0.8864	<0.0001**	0.7302 / <0.0001**	0.6067
Lateral Bending	0.3192	0.0056*	0.0250* / 0.0460*	0.0171*
Flexion-Extension	0.7865	<0.0001**	<0.0001** / 0.0668	0.8946
Resultant Kinematics	0.4891	<0.0001**	<0.0001** / 0.0019*	0.0428*

Note: * Statistically significant with *p*-value < 0.05
** Statistically significant with *p*-value < 0.0001

of TRP, across all participants. Furthermore, Table 2 provides information regarding the p-values for significant levels (slip and trip) of condition. Condition was found to be statistically significant (p-value < 0.05) for all three angles of TRP for all participants. Values of resultant kinematics for the slip and trip trials were found statistically significant (p-value < 0.05) for all participants

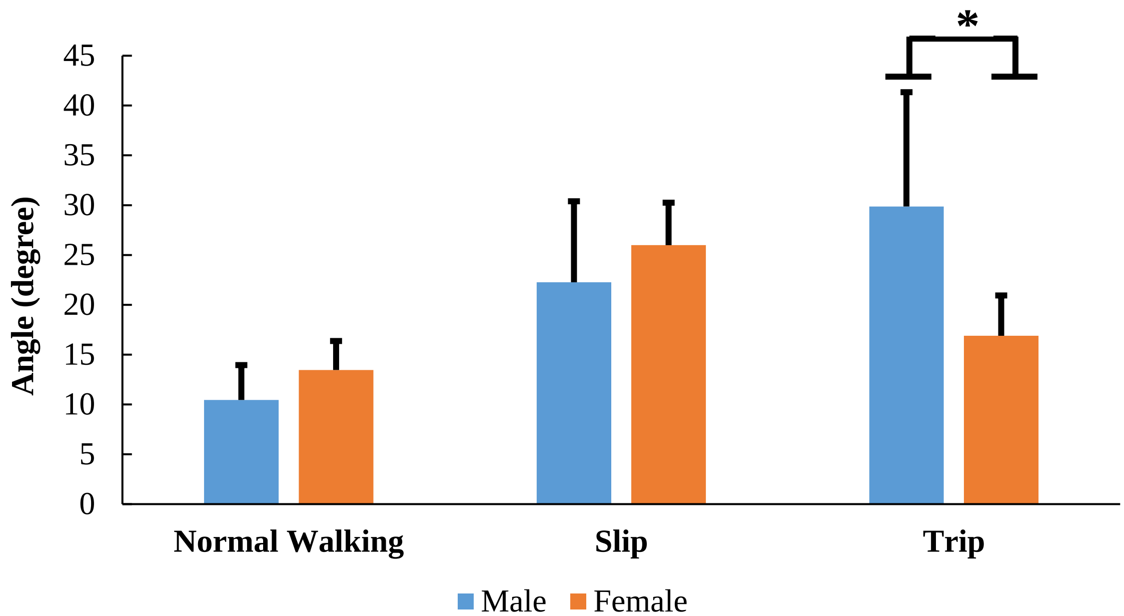

Figure 15: Two factor interaction of gender and condition for the range of lateral bending motion.

Note: * The range of lateral bending motion for trip in females was statistically significant.

compared to normal walking. However, for slip trials, the angle of TRP in axial rotation and for trip trials, the angle of TRP in flexion-extension did not observed any significant effects compared to normal walking. Gender was not associated with any statistically significant values for the resultant kinematics. But, for the angle of TRP in lateral bending and the resultant kinematics, the two-factor interaction of gender and condition was found statistically significant (Figure 15). However, no specific cause was determined for this finding.

Figure 16 represents a statistical summary of the mean of the resultant motion of TRP (degree) across the different participants during normal walking, slip, and trip trials. Based on post-hoc analysis (using Tukey's HSD), conditions not connected by the same letter are significantly different with a p-value of <0.0001 for the slip trial and 0.0019 for the trip trial.

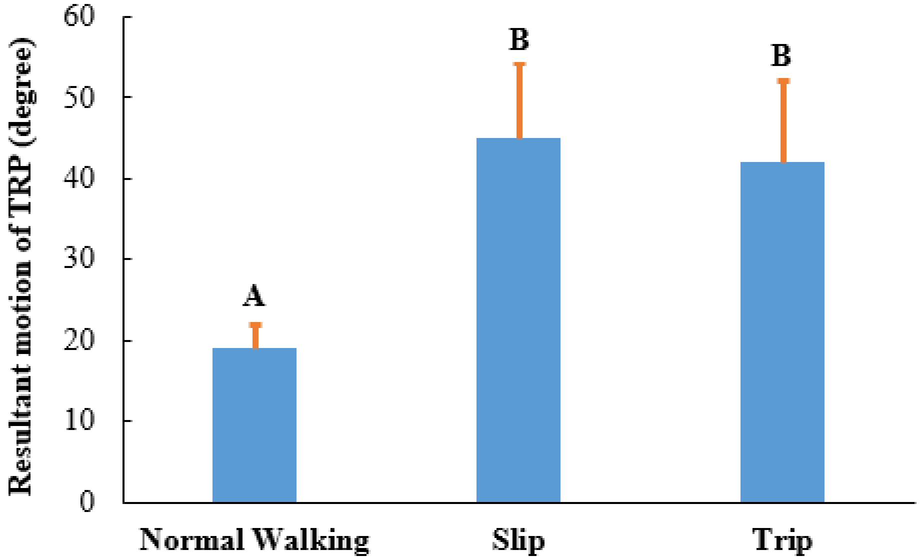

Figure 16: Statistical summary (post-hoc) of resultant motion of TRP (degree) across the different participants during normal walking, slip and trip trials.

Note: Conditions (normal walking, slip, trip) not connected by the same letter are significantly different with a p-value of < 0.05.

Hence, our results confirm the hypothesis that lumbar kinematics would increase significantly during slip and trip trials compared to normal walking.

3.2. Lumbar Muscle Activity

Lumbar muscle activity increased significantly for all participants during the slip and trip trials compared to normal walking. Lumbar muscle activity was recorded for the fourteen bilateral flexor and extensor muscles around the lumbar and the belly region, which are classified as follows:

Flexors: Internal oblique (IO), rectus abdominis (RA), external oblique (EO).

Extensors: Multifidus (MF), longissimus thoracis pars lumborum (LTL), iliocostalis lumborum pars lumborum (ILL), longissimus thoracis pars thoracis (LTT).

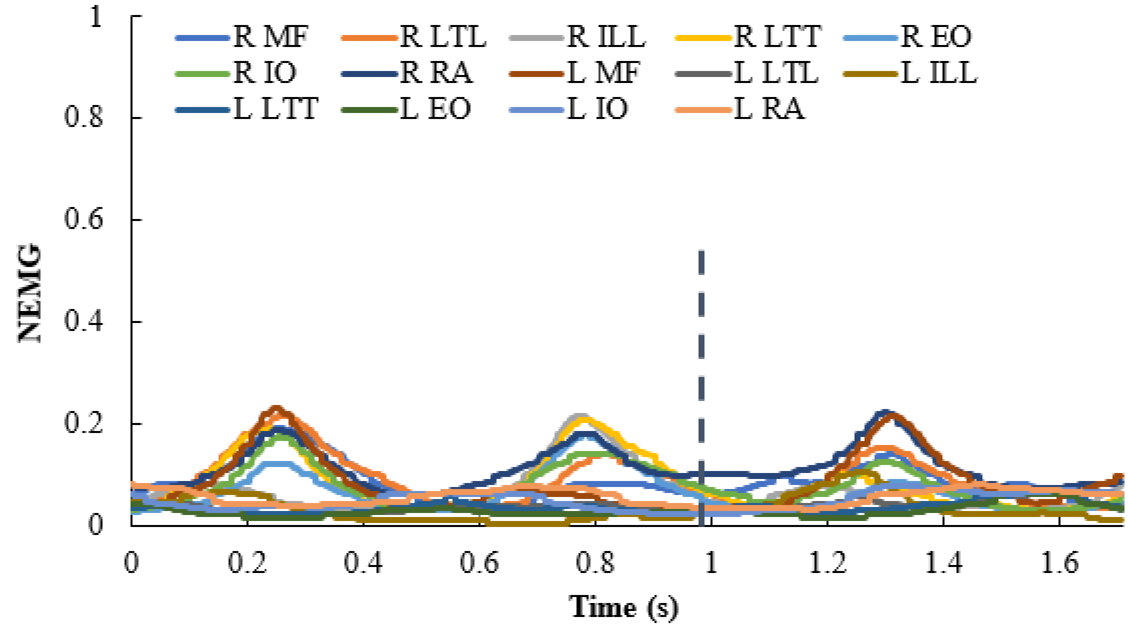

Figure 17: Normalized muscle activity (NEMG) for bilateral (right & left) lumbar muscles during normal walking (multifidus (MF), longissimus thoracis pars lumborum (LTL), iliocostalis lumborum pars lumborum (ILL), longissimus thoracis pars thoracis (LTT), external oblique (EO), internal oblique (IO), rectus abdominis (RA)). Dotted line indicates the heel contact on the force plate at 0.98 seconds.

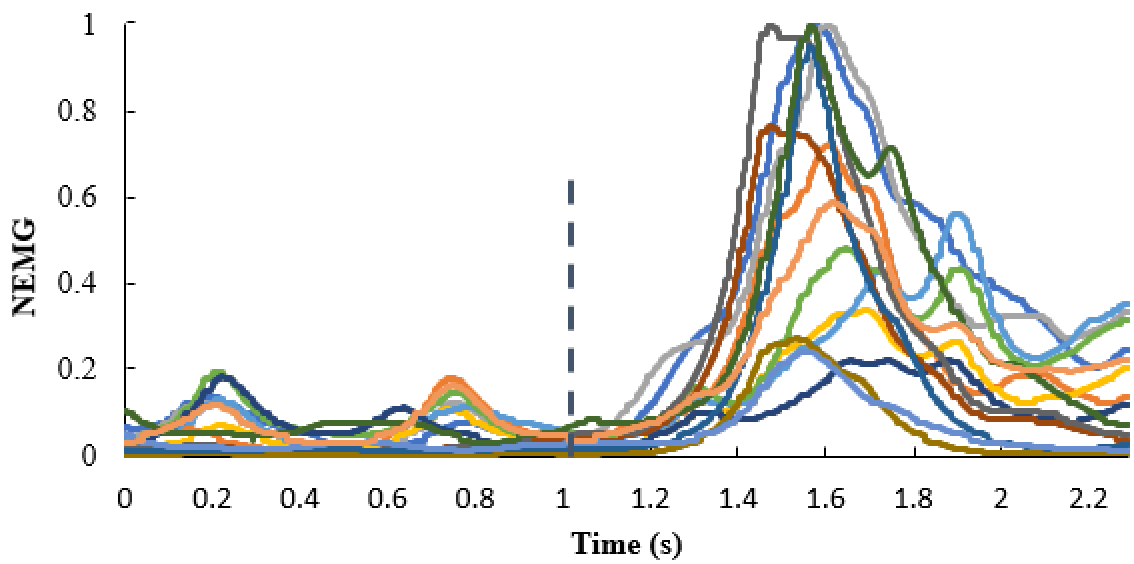

Figure 18: Normalized muscle activity (NEMG) for bilateral (right & left) lumbar muscles during the slip trial (multifidus (MF), longissimus thoracis pars lumborum (LTL), iliocostalis lumborum pars lumborum (ILL), longissimus thoracis pars thoracis (LTT), external oblique (EO), internal oblique (IO), rectus abdominis (RA)). Dotted line indicates the heel contact on the force plate at 1.02 seconds.

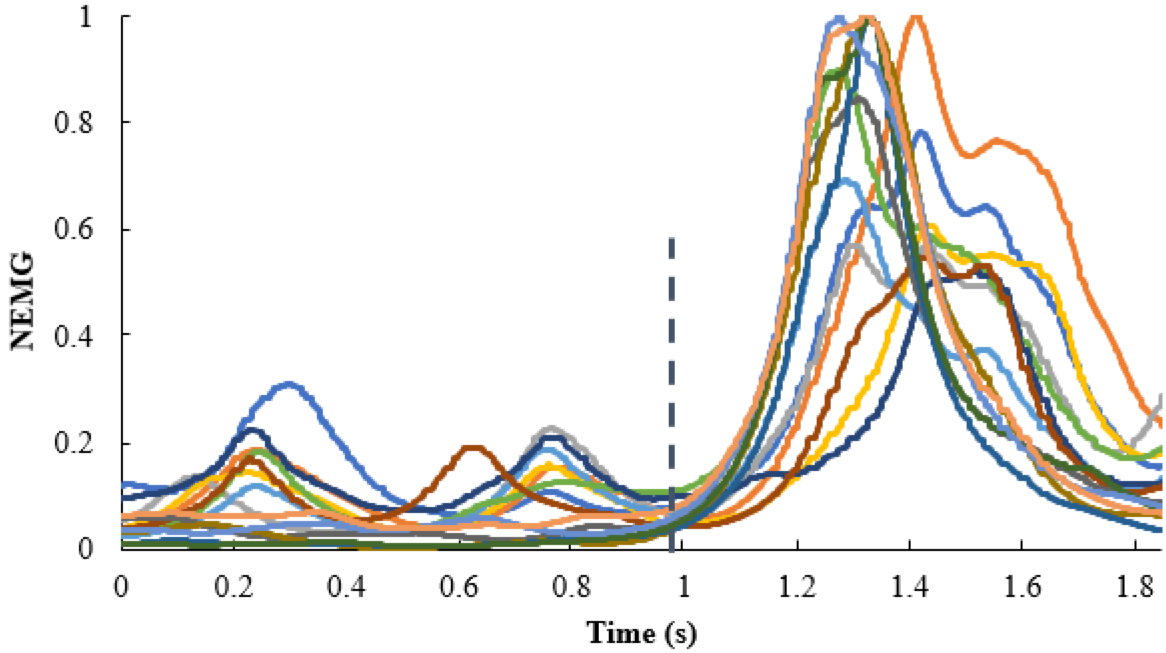

Figure 19: Normalized muscle activity (NEMG) for bilateral (right & left) lumbar muscles during the trip trial (multifidus (MF), longissimus thoracis pars lumborum (LTL), iliocostalis lumborum pars lumborum (ILL), longissimus thoracis pars thoracis (LTT), external oblique (EO), internal oblique (IO), rectus abdominis (RA)). Dotted line indicates the heel contact on the force plate at 0.95 seconds.

Figures 17,18 &19 show NEMG values calculated during normal walking, slip, and trip trials for one of the participants, respectively. Muscle activity was normalized with respect to the recorded MVC data during the first phase of the experimental trial. As seen in Figure 17**,** for normal walking the NEMG follows a constant profile and the range of values is similar before and after the heel strike throughout the gait.

For the slip trial, all fourteen muscles demonstrated a substantial increase in activity, with muscles reaching their maximum activation levels around 0.5 seconds after the right-foot heel strike at 1.02 seconds (Figure 18). And for the trip trial, again, all fourteen muscles demonstrated

Table 3: Mean (SD) of the maximum normalized muscle activity (NEMG) of bilateral lumbar flexor (IO EO RA) and extensor muscles (MF ILL LTL LTT) during normal walking, slip, and trip trials across the different participants.

Muscle Type	Normal Walking Mean (SD)	Slip Trial Mean (SD)	Trip Trial Mean (SD)
R MF	0.265 (0.176)	0.683 (0.271)	0.673 (0.254)
R LTL	0.317 (0.186)	0.83 (0.217)	0.73 (0.274)
R ILL	0.192 (0.115)	0.701 (0.297)	0.567 (0.252)
R LTT	0.202 (0.098)	0.757 (0.228)	0.743 (0.261)
R EO	0.235 (0.106)	0.592 (0.307)	0.671 (0.284)
R IO	0.237 (0.132)	0.645 (0.336)	0.75 (0.234)
R RA	0.204 (0.104)	0.596 (0.352)	0.696 (0.252)
L MF	0.227 (0.139)	0.555 (0.293)	0.579 (0.24)
L LTL	0.185 (0.123)	0.784 (0.218)	0.763 (0.208)
L ILL	0.16 (0.077)	0.75 (0.259)	0.82 (0.272)
L LTT	0.0723 (0.113)	0.627 (0.298)	0.572 (0.266)
L EO	0.063 (0.069)	0.708 (0.29)	0.573 (0.29)
L IO	0.211 (0.126)	0.756 (0.275)	0.808 (0.188)
L RA	0.152 (0.095)	0.728 (0.26)	0.725 (0.27)
Mean NEMG	0.195 (0.088)	0.694 (0.182)	0.691 (0.182)

a significant increase in activity; specifically, roughly 0.3 seconds after the right-foot heel strike at 0.95 seconds, the muscles reached their maximum activation levels (Figure 19). All fourteen muscles were activated with a similar pattern during the slip and trip trials. More than a three-fold increase in NEMG values was observed for the slip and trip trials compared to normal walking.

In general, muscle activation levels were substantially higher for all participants during the slip and trip trials compared to normal walking. Table 3 provides a detailed summary regarding the mean (SD) of maximum muscle activity for each of the fourteen muscles for all participants during normal walking, slip, and trip trials. Across all participants, the mean of the maximum

Table 4: Summary of ANOVA results for the main and first-order interaction effect of gender and condition on normalized muscle activity (NEMG).

Muscle Type	Gender	Condition	Slip / Trip	Gender*Condition Interaction
R MF	0.5957	<0.0001**	0.0003*	0.7604
			0.0008*	
R LTL	0.4546	<0.0001**	<0.0001**	0.1517
			0.0094*	
R ILL	0.7030	<0.0001**	<0.0001**	0.8044
			0.1726	
R LTT	0.6410	<0.0001**	0.0001*	0.0859
			<0.0001**	
R EO	0.7875	<0.0001**	0.0200*	0.6528
			0.0002*	
R IO	0.3117	<0.0001**	0.031*	0.9699
			0.0001*	
R RA	0.6139	<0.0001**	0.0371*	0.7224
			<0.0001**	
L MF	0.4515	<0.0001**	0.0332*	0.08963
			0.0046*	
L LTL	0.6020	<0.0001**	<0.0001**	0.5375
			0.0001*	
L ILL	0.7492	<0.0001**	0.0026*	0.9287
			<0.0001**	
L LTT	0.3863	<0.0001**	0.0009*	0.8405
			0.0081*	
L EO	0.8127	<0.0001**	<0.0001**	0.9695
			0.0153*	
L IO	0.6736	<0.0001**	0.0032*	0.2948
			0.0001*	
L RA	0.6107	<0.0001**	0.0004*	0.5967
			0.0004*	
Mean NEMG	0.7359	<0.0001**	<0.0001**	0.5686
			<0.0001**	

Note: * Statistically significant with *p*-value < 0.05
**** Statistically significant with *p*-value < 0.0001**

NEMG for the slip and trip trials showed more than a threefold increase compared to the normal walking. Table 4 provides a summary of the results from repeated measure ANOVAs of NEMG, including all fourteen muscles along with the mean NEMG and the p-values for significant levels (slip and trip) of condition. For all participants, condition was found to be highly significant (p-value < 0.0001) for all fourteen muscles. The NEMG values for the slip and trip trials were found statistically significant (p-value < 0.05) for all participants compared to normal walking. Moreover, it can be noted neither gender - nor the two-factor interaction of gender and condition - displayed any statistically significant effect on the NEMG.

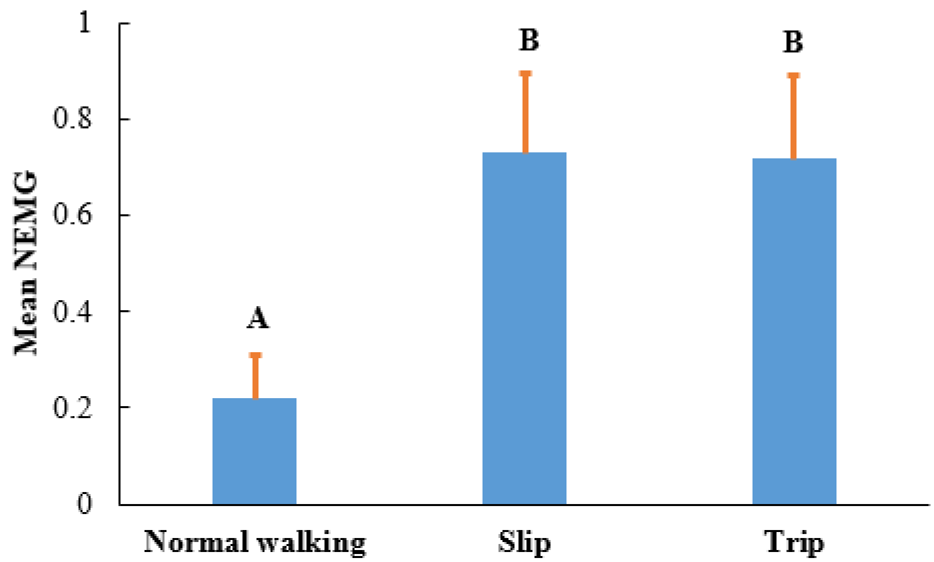

Figure 20: Statistical summary (post-hoc) of mean of maximum NEMG of the bilateral flexor and extensor muscles across the different participants during normal walking, slip and trip trials.

Note: Conditions (normal walking, slip, trip) not connected by the same letter are significantly different with a p-value of < 0.0001

Figure 20 represents a statistical summary of the mean of maximum NEMG of the combined bilateral flexor and extensor muscles across the different participants during normal walking, slip, and trip trials. Based on the post-hoc analysis (using Tukey's HSD), conditions not connected by the same letter are significantly different with a p-value of <0.0001.

3.3. Lumbosacral Reaction Forces and Moments

The obtained lumbar kinematics and NEMG were used as an input to the EMG-based lumbar spine model, which provided the lumbosacral reaction forces and moments. The EMG-based model of the lumbar spine quantified the lumbosacral forces and moments for all the trials, including normal walking, slip, and trip. As hypothesized earlier, the lumbosacral forces and moments increased significantly during the slip and trip trials compared to normal walking.

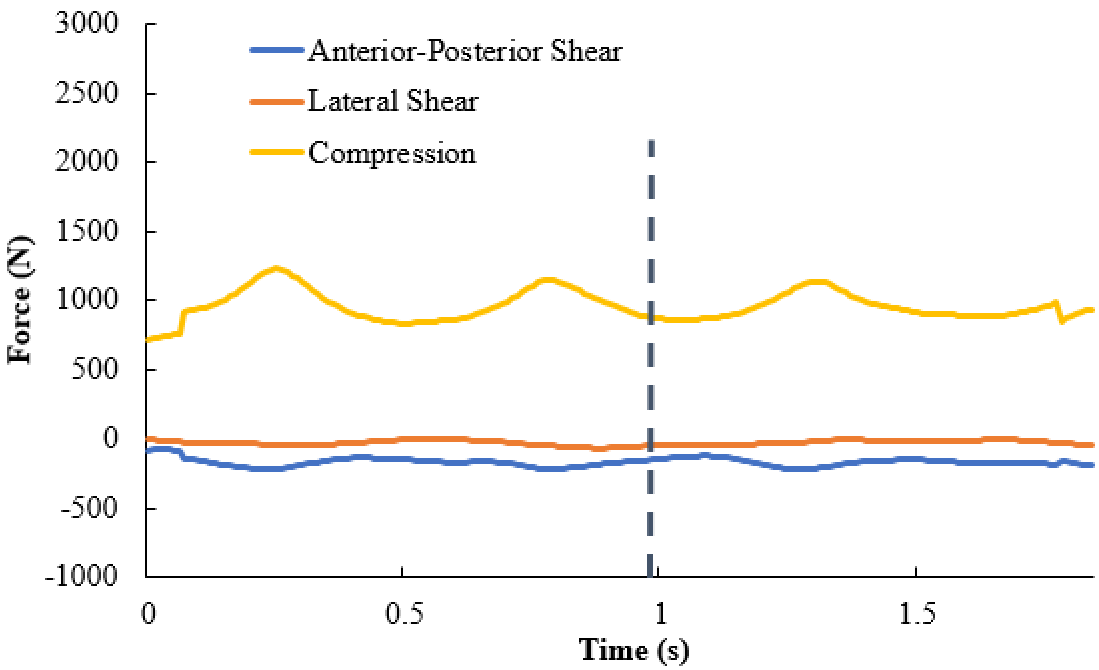

Figure 21: Lumbosacral forces (anterior-posterior shear, lateral shear, compression) during normal walking. Dotted line indicates the heel contact on the force plate at 0.98 seconds.

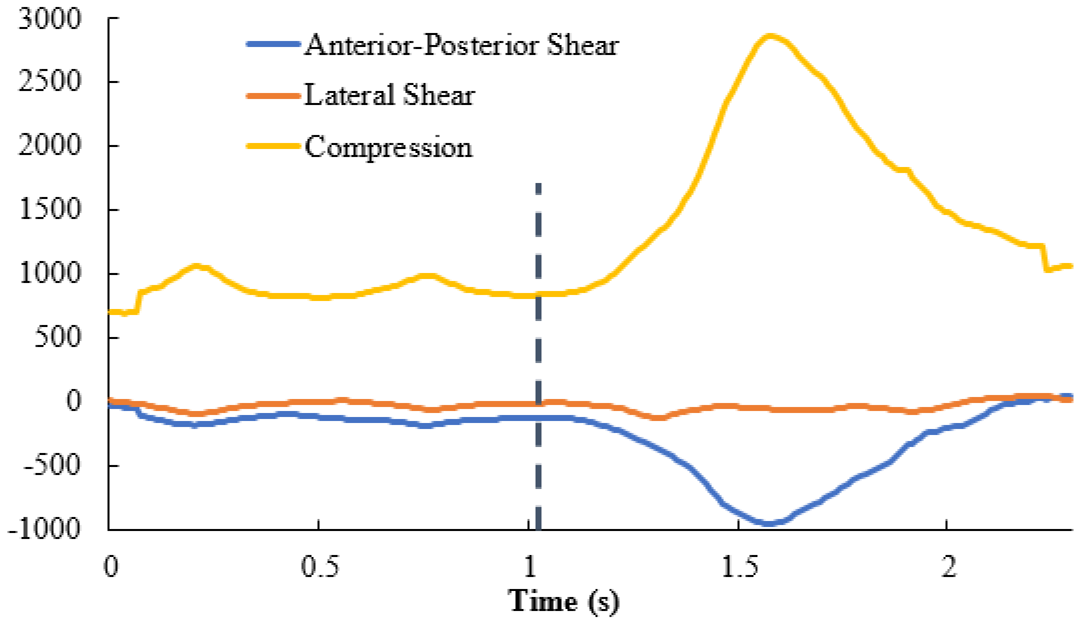

Figure 22: Lumbosacral forces (anterior-posterior shear, lateral shear, compression) during the slip trial. Dotted line indicates the heel contact on the force plate at 1.02 seconds.

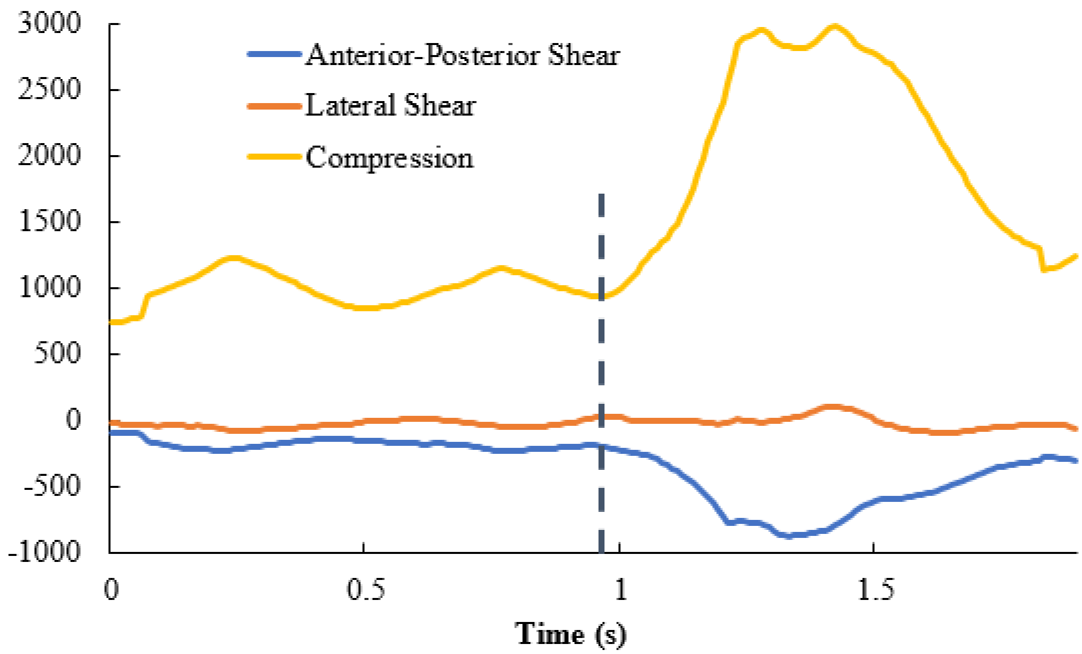

Figure 23: Lumbosacral forces (anterior-posterior shear, lateral shear, compression) during the trip trial. Dotted line indicates the heel contact on the force plate at 0.95 seconds.

Lumbosacral forces were measured in three different directions: anterior-posterior shear, lateral shear, and compression. During normal walking (Figure 21), all the three forces showed nearly similar values before and after the right-foot heel strike represented by the dashed line. In contrast, for the slip and trip trials, the three forces increased significantly after the right-foot heel strike (Figure 22 & 23). For the slip and trip trials, the L5/S1 compression force showed an increase to 2870 N and 2980 N respectively, compared to 1220 N for normal walking. Similarly, the anterior-posterior shear force for the slip and trip trials increased to 960 N and 850 N, respectively, compared to 220 N for normal walking. We also noted that the lateral shear force increased to 135 N and 110 N during the slip and trip trials versus 40 N for normal walking. The increase in the tri-axial forces during the slip and trip trials was observed at roughly 0.5 seconds after the right-foot heel strike.

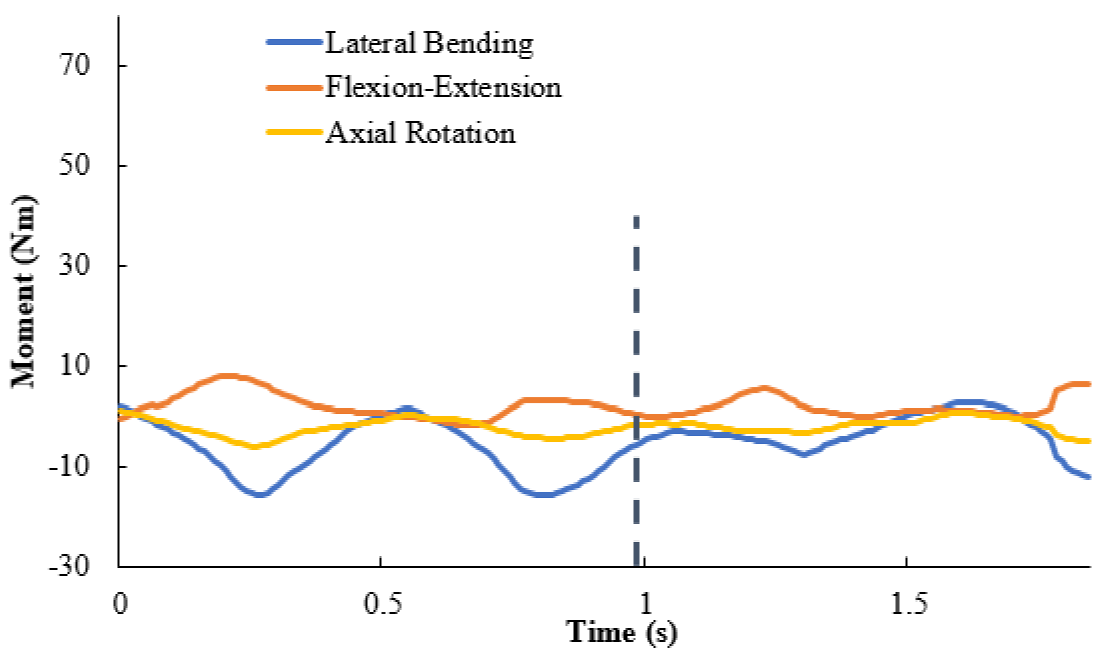

Figure 24: Lumbosacral moments (lateral bending, flexion-extension, axial rotation) for normal walking. Dotted line indicates the heel contact on the force plate at 0.98 seconds.

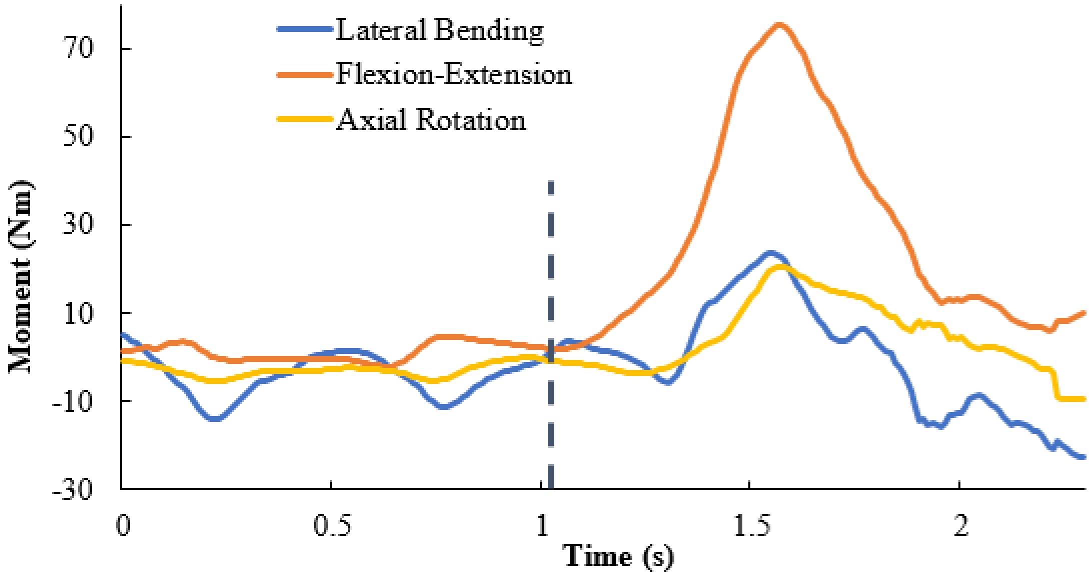

Figure 25: Lumbosacral moments (lateral bending, flexion-extension, axial rotation) during the slip trial. Dotted line indicates the heel contact on the force plate at 1.02 seconds.

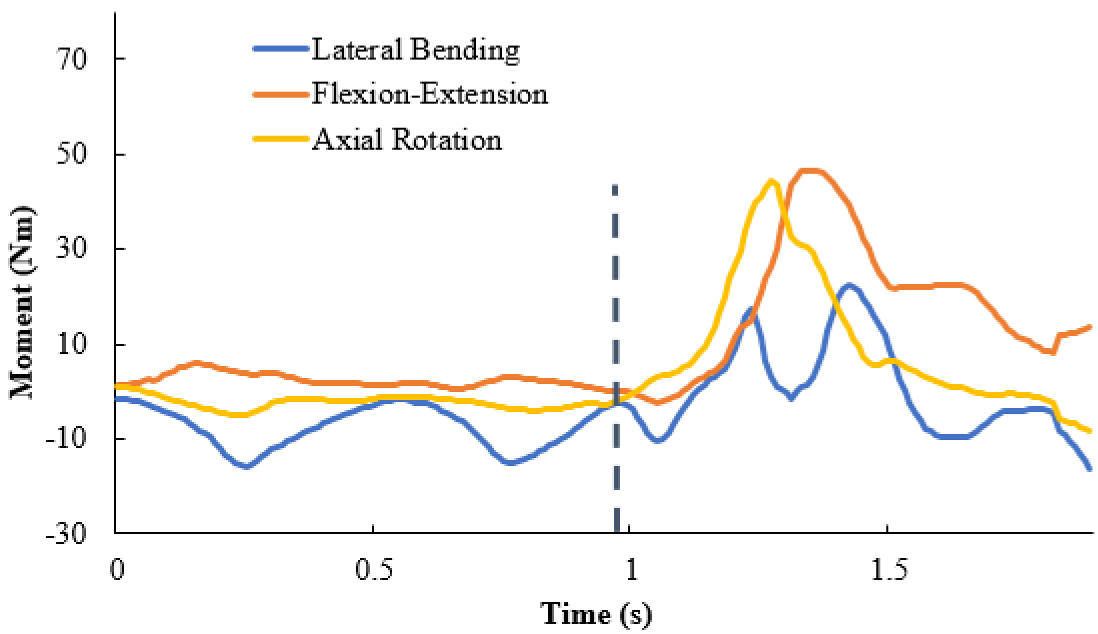

Figure 26: Lumbosacral moments (lateral bending, flexion-extension, axial rotation) during the trip trial. Dotted line indicates the heel contact on the force plate at 0.95 seconds.

All L5/S1 reaction moments were recorded in three different directions of rotation: lateral bending, flexion-extension, and axial rotation. As noted earlier with our force data, during normal walking (Figure 24) all three moments recorded a similar profile before and after the heel strike, with values hovering around zero. Conversely, for the slip and trip trials, a substantial increase in the moments was observed after the right-foot heel strike (Figure 25 & 26). The flexion-extension moment for the slip and trip trials increased to 75 Nm and 46 Nm, respectively, compared to 8 Nm for normal walking. The lateral bending moment showed an increase from 15 Nm for normal walking to 24 Nm and 23 Nm for the slip and trip trials. Similarly, the axial rotation moment increased to 21 Nm and 44 Nm for the slip and trip trials compared to 6 Nm for normal walking. A substantial increase in the moments occurred roughly 0.5 seconds after the right-foot heel strike.

Table 5: Mean (SD) of the maximum lumbosacral forces & moments during normal walking, slip, and trip trials across the different participants.

Forces & Moments	**Normal Walking Mean (SD)**	**Slip Trial Mean (SD)**	**Trip Trial Mean (SD)**
A/P Shear Force (N)	230.79 (55.6)	929.83 (184.86)	798.52 (280.48)
Lateral Shear Force (N)	72.13 (32)	155.81 (81.03)	220.06 (160.44)
Compression Force (N)	1148.3 (213)	2780.25 (486.64)	2769.43 (599.10)
Lateral Bending Moment (Nm)	11.64 (5.72)	31.62 (10.1)	30.81 (12.22)
Flexion-Extension Moment (Nm)	7.43 (3.18)	35.83 (19.08)	28.4 (14.25)
Axial Rotation Moment (Nm)	5.11 (2.18)	14.83 (8.45)	22.31 (8.8)
Resultant Force (N)	1173.96 (219.68)	2938.68 (508.56)	2898.4 (643.71)
Resultant Moment (Nm)	15.59 (4.36)	51.80 (18.43)	50.12 (12.14)

Table 6: Summary of ANOVA results for the main and first-order interaction effect of gender and condition on lumbosacral forces and moments.

Forces and moments	Gender	Condition	Slip / Trip	Gender*Condition Interaction
A/P Shear Force	0.7299	<0.0001**	<0.0001**	0.8669
			0.0025*	
Lateral Shear Force	0.4377	0.0056*	0.7630	0.5579
			0.0071*	
Compression Force	0.3405	<0.0001**	<0.0001**	0.2439
			<0.0001**	
Lateral Bending Moment	0.2689	0.0002*	0.0151*	0.8650
			0.0249*	
Flexion-Extension Moment	0.0978	<0.0001**	0.0013*	0.5692
			0.1925	
Axial Rotation Moment	0.7469	<0.0001**	0.7015	0.3721
			<0.0001**	
Resultant Force	0.3720	<0.0001**	<0.0001**	0.3020
			<0.0001**	
Resultant Moment	0.4909	<0.0001**	0.0005*	0.5104
			0.0020*	

Note: * Statistically significant with *p*-value < 0.05
**** Statistically significant with *p*-value < 0.0001**

In general, all the participants showed a substantial increase in lumbosacral forces and moments during unexpected walkway perturbations (i.e., during slips and trips) compared to normal walking. Table 5 provides a detailed summary regarding the mean (SD) of the maximum forces and moments for all the participants during normal walking, slip, and trip trials. Summary results from repeated measure ANOVAs of lumbosacral forces and moments are represented in Table 6. For all participants, condition was found to be statistically significant (p-value < 0.05) for the individual forces (A/P shear, lateral shear, and compression) and moments (lateral bending, flexion-extension, and axial rotation), as well as the resultant force and moment. Furthermore, detailed information pertaining to the p-values for significant levels (slip and trip) of the condition are presented in Table 6. Note also that lumbosacral loads increased significantly (p-value < 0.05) for the slip and trip trials compared to normal walking. Neither gender nor the two-factor interaction of gender and condition displayed any significant effect on lumbosacral forces and moments.

A statistical summary of the mean of the resultant forces and moments across the different participants during normal walking, slip, and trip trials is represented in Figures 27 and 28. Based on the post-hoc analysis (using Tukey's HSD), conditions not connected by the same letter are significantly different with a p-value of <0.0001. Hence, this data supports our hypothesis that lumbosacral forces and moments would significantly increase during slip and trip trials compared to normal walking.

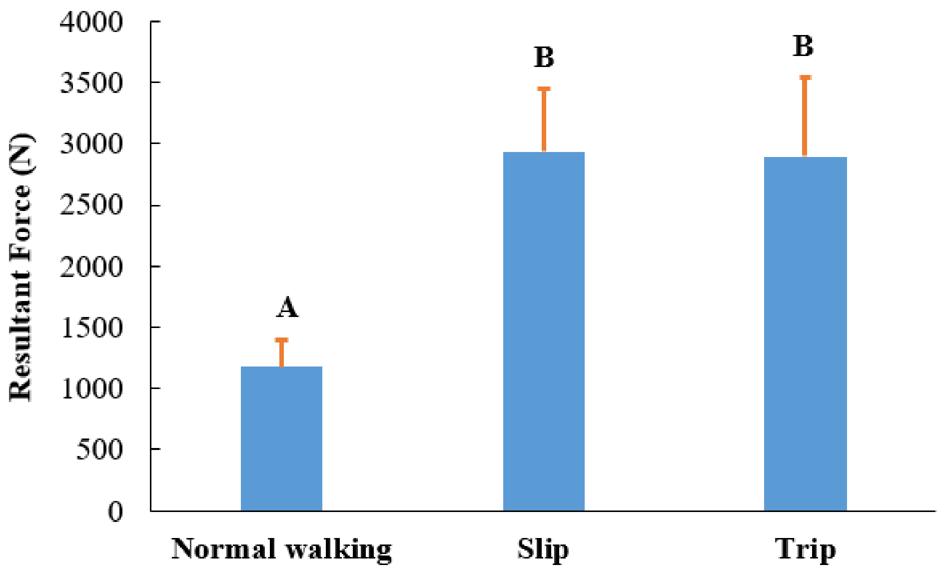

Figure 27: Statistical summary (post-hoc) of resultant force across the different participants during normal walking, slip and trip trials.

Note: Conditions (normal walking, slip, trip) not connected by the same letter are significantly different with a p-value of < 0.0001

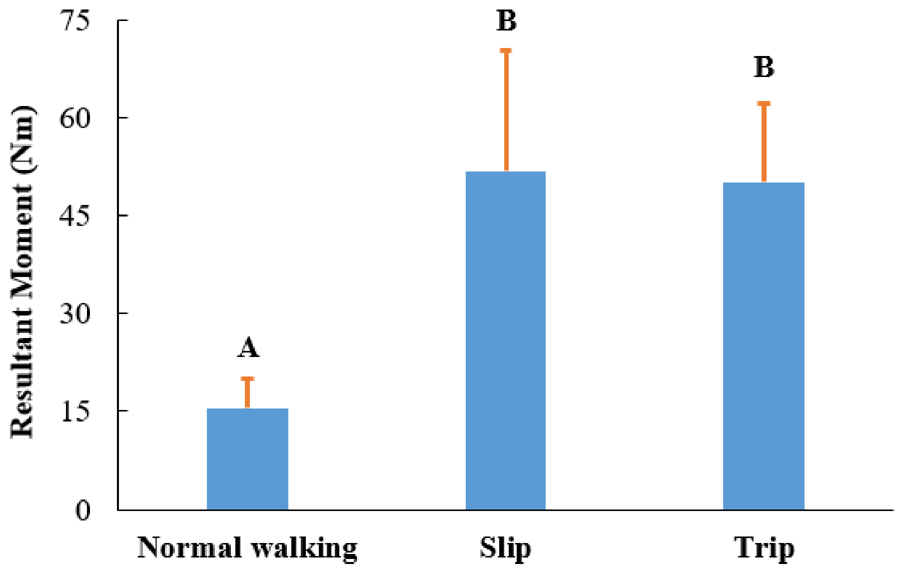

Figure 28: Statistical summary (post-hoc) of resultant moment across the different participants during normal walking, slip and trip trials.

Note: Conditions (normal walking, slip, trip) not connected by the same letter are significantly different with a p-value of < 0.0001

3.4.Inverse Dynamics

For validation purpose, L5/S1 reaction moments from the EMG-based spine model were compared to a 3D, inverse dynamics model developed in Visual3D. However, the inverse dynamics model was unable to successfully validate the moments obtained from the EMG-based spine model because of the quality of the raw experimental data. Important key limiting factors associated with this study are addressed below:

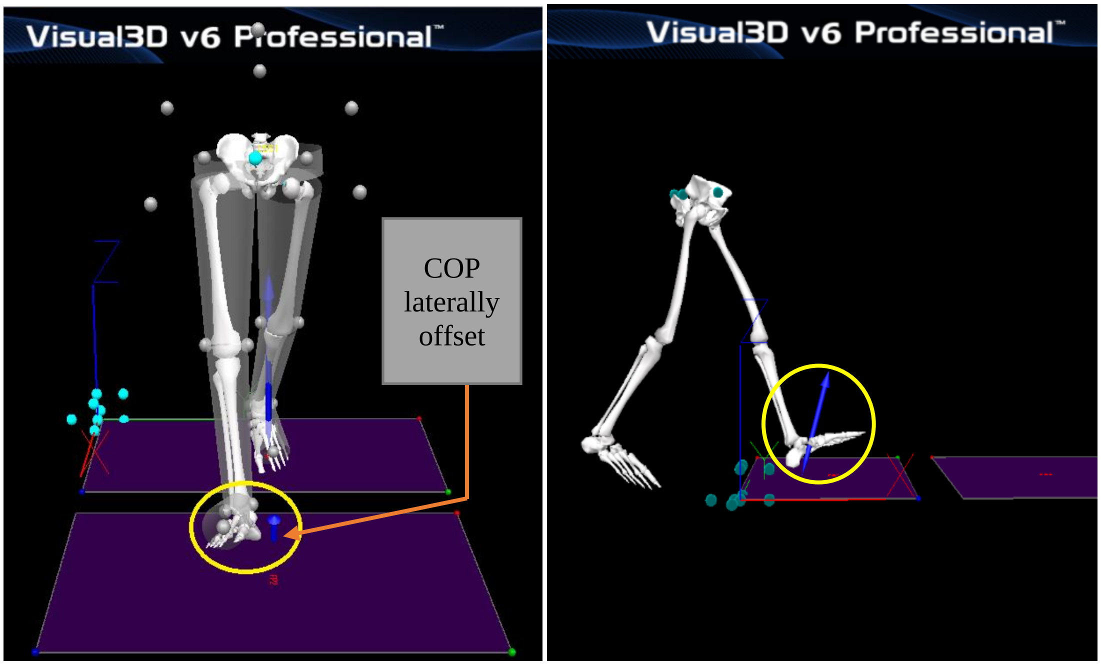

Figure 29: Ground reaction force from different views in our Visual3D model based on the raw experimental data.

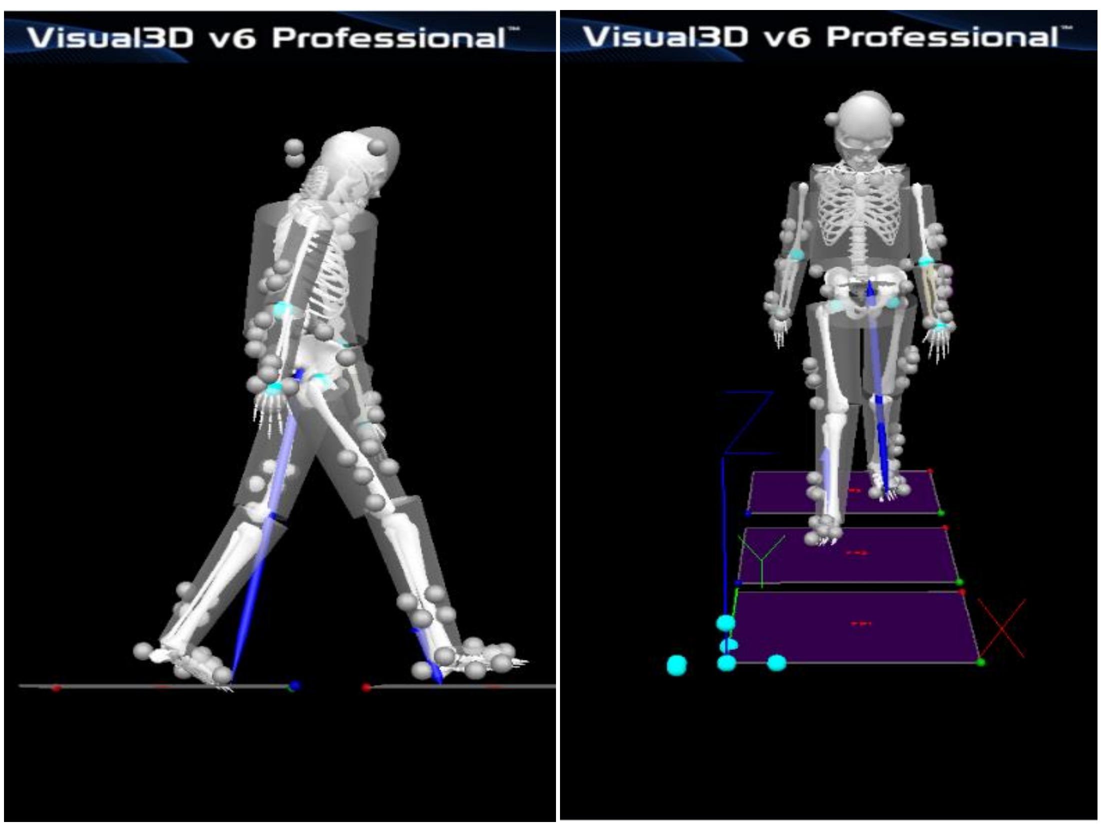

Figure 30: Ground reaction force from different views in sample model for normal walking provided by C-motion.

- The ground reaction force (GRF) from the force-plate is shown in Figure 29 in different planes. During the heel-strike phase, the direction of GRF is not aligned towards the trunk (L5/S1) and instead points in a different direction (represented by the yellow circles in Figure 29). This factor could have created a larger moment arm that resulted in a substantial increase in the moment at L5/S1. The results were compared with a sample model for normal walking provided by C-motion shown in Figure 30. Here, the direction of GRF is properly aligned towards L5/S1 during the entire gait cycle.

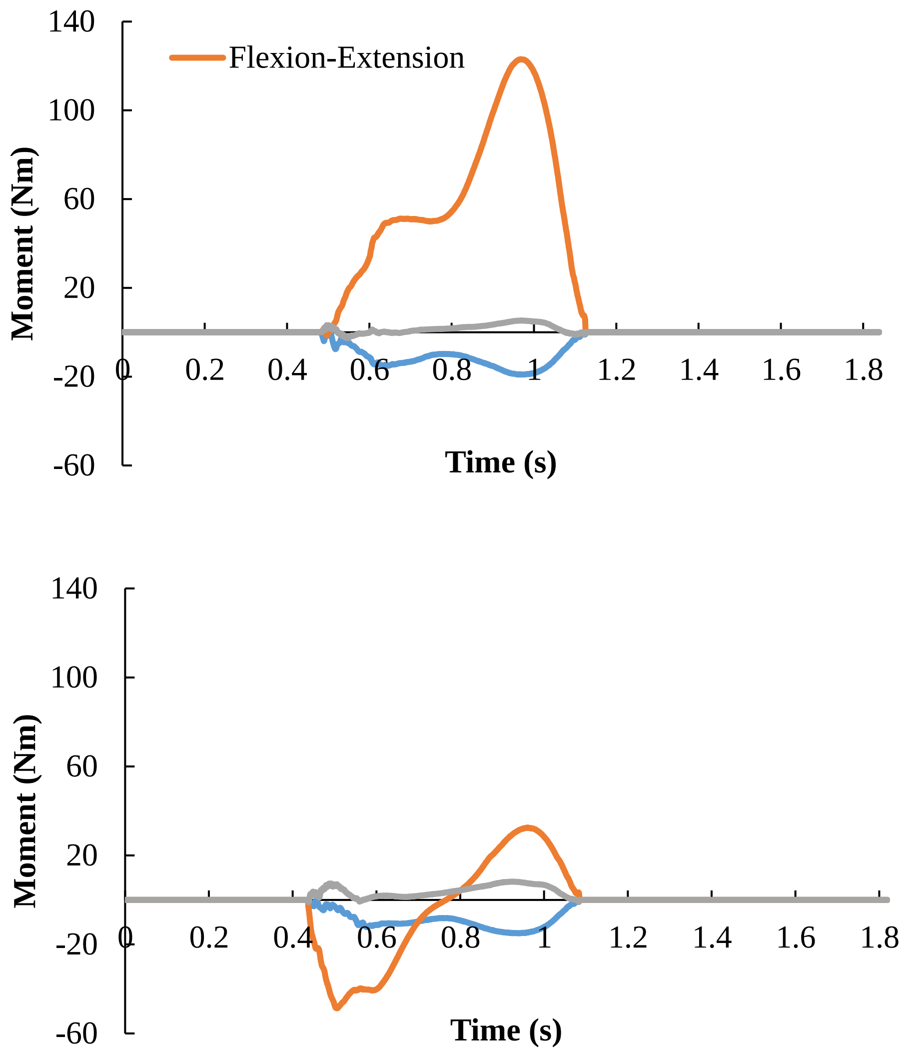

Figure 31: Flexion-extension moment from the raw force-plate data for two different normal walking trials for the same participant.

- Figure 29 (the left side) indicates that the position of the COP is laterally offset by some distance, instead of having the COP beneath the foot. This is yet another possible source of error that might have impacted the moment calculations from the inverse dynamics model.
- The raw data for moment on force-plate seems to be a concern that could have affected the inverse dynamics outcome. For example, Figure 31 shows the moment from raw data for two different normal walking trials for the same participant. The moment in flexion-extension has a very different profile for both trials for the same participant. This factor might have affected the final output of inverse dynamics because of the variability in the raw flexion-extension moment.

But, we’re able to qualitatively verify the obtained moments from EMG-based lumbosacral model with the predefined model in Visual3D for normal walking (reference). The respective moments from the predefined normal walking model in Visual3D for flexion-extension, axial rotation, and lateral bending were 11, 4 & 14 Nm which is similar to the values reported in the current study (Table 6).

4. Discussion

There is a wealth of scholarly reports investigating a range of essential parameters that occur during normal human walking, including lumbar kinematics, muscle activity, and low back loads (e.g., Murray et al., 1984; Cappozzo, 1983, 1984; Rowe & White, 1996). Cham and Redfern (2001) focused on the forces and joint reaction moments acting on the lower extremity during normal gait, while other researchers have investigated lumbar vertebral compressive loads during normal walking (e.g., Cappozzo, 1984; Callaghan, Patla, & McGill, 1999). The current experimental study adds to the literature by adopting a unique approach for analyzing low back loads, lumbar kinematics, and lumbar muscle activity during unanticipated and unexpected walkway perturbations that result in slips and trips - but not falls. Prior to undertaking this investigation, we hypothesized that lumbar kinematics, lumbosacral loads, and muscle activity would increase during walking perturbations such as slip and trip in comparison to normal walking. During normal walking, experimental results indicated low levels of lumbar kinematics, lumbar muscle activity, and moderate loading on the spine. In contrast, unexpected slip - and trip-induced walkway perturbations resulted in a significant increase in kinematics, muscle activity, and lumbosacral forces and moments. No statistical significance was reported in terms of relationships between age, height, weight, gender and the following response variables: lumbar kinematics, lumbar muscle activity, lumbosacral forces and moments.

4.1. Lumbar Kinematics

The motion of the lumbar spine (including the trunk and pelvis) was analyzed in three different directions of rotation: flexion/extension, axial rotation, and lateral bending. As discussed earlier in the Results section, the motion of the trunk relative to the pelvis (TRP) showed a significant increase during the slip- and trip-induced trials compared to normal walking (Figures

12, 13 and 14). For normal walking, the motion of the TRP exhibited consistent patterns within, as well as between, participants. The flexion-extension motion of the TRP demonstrated a relatively small range of motion, whereas both the axial rotation and the lateral bending motion of the TRP observed a pattern with three peaks, since we collected data for the three steps around the force platforms (Figure 12). At the beginning of the gait, following the right-foot heel contact, there was a right lateral flexion of the spine towards the side (Figure 13). Then, the spine underwent a contralateral flexion (i.e. to the left side) towards the end of the right foot toe-off, followed by a right lateral flexion at the right foot heel strike. For the axial twist (axial rotation), upon the right-foot heel strike, the motion of the TRP observed a twist towards the right side (clockwise direction). After the right foot toe-off, the motion of the TRP showed a steady rotation towards the counter-clockwise direction, followed by a right twist again at the right-foot heel strike (Figure 14). We observed that the motion of the TRP occurred mainly due to the motion of the pelvis. The lateral bending motion of the spine experienced three peaks that occurred at close to heel-strike events, while axial twisting experienced three peaks roughly 100 milliseconds (ms) after the heel-strike events. The range of angles for the motion of the TRP for normal walking was found to be consistent with earlier studies. Moreover, Callaghan and coworkers (1999) reported the range of the lumbar spine motion relative to the pelvis as: 2.72-10.25° for flexion-extension, 1.12-7.13° for lateral bending and 3.52-14.69° for axial rotation. In the current study, the range of motion of the TRP during normal walking indicated quite similar values: 4.13-7.01° for flexion-extension, 8.52-15.4° for lateral bending and 10.7-16.08° for axial rotation.

For the slip and trip trials, the motion of the TRP increased substantially in all three directions of rotation (flexion/extension, axial rotation, and lateral bending) with more than a two-fold increase in angle values. The increase in the motion of the spine was observed roughly 250 ms after the heel strike on the force plate, which is consistent with values reported by Cham and

Redfern (2001) who indicated that the motion of the spine occurred on average 190-250 ms after heel strike

4.2. Lumbar Muscle Activity

During normal walking, the NEMG for all flexor and extensor muscles recorded a constant profile with three peaks observed during the gait cycle (Figure 17). The three peaks occurred roughly 200 ms after each of the three heel strikes during normal gait. No discernible or irregular patterns were reported for NEMG across the different participants during normal walking. According to the literature, the increase in muscle activity after a heel strike during normal walking is intended to counter the flex movement of the trunk (Waters & Morris, 1972). To maintain balance while walking, the trunk must position and balance itself on the pelvis, which moves along vertical, lateral and rotational axes. During normal walking, due to the lateral movement made by the trunk to position itself over the supporting foot, the lumbar spine muscles are then activated to provide lateral stability to the trunk. Based on a summary of activation levels across the different participants during normal walking, the R LTL muscle exhibited the highest activation with a mean (SD) of 0.317 (0.188); in contrast, the L EO muscle was found to be the least active with a mean of 0.063 (0.069). Overall, the mean (SD) of the maximum NEMG across all participants during normal walking was found to be 0.195 (0.088). Low-back muscle activity during normal walking observed in the current study tends to support the existing literature. As an example, Murray and coworkers (1984) reported the NEMG value for spine muscles during normal walking to be 0.27 (0.07).

For the slip and trip trials, all the flexor and extensor muscles demonstrated a sharp increase in muscle-activation levels roughly 0.4 second after heel strike (Figures 18 and 19). For both the slip and trip trials, the mean (SD) values of the maximum NEMG increased almost four times compared to normal walking (Table 1). In terms of specific sets of muscles, the mean activation levels were similar for both the left- and the right-side muscles during perturbed trials (i.e. slips and trips). Since slips and trips tend to occur unexpectedly, the body movements we recorded for the different participants in this study were very dynamic and varied among the individuals. As a result, no specific patterns were observed in the muscle activity of different muscles during slip and trip trials. Based on an examination of the activation levels across the different participants during the slip trials, the R LTL muscle showed highest activation, with a mean (SD) value for the maximum NEMG as 0.83 (0.217). Conversely, for the trip trials, the L ILL muscle experienced the highest activation level with a mean (SD) value for the maximum NEMG as 0.82 (0.272). During the slip and trip trials, some muscles exceeded maximum muscle activity during the MVC trials. In such cases, the value of NEMG was capped to 1.0. It should be noted, however, that there was no consistent pattern observed in terms of any specific NEMG exceeding the value of 1.0 during the slip and trip trials. This outcome might be due to the fact that, as noted above, the bodily response varies among individuals due to the dynamic nature of slips and trips as a result of walkway perturbations.

Such high muscle activation levels during the slip- and trip-induced trials could be associated with the muscle force generation required to regain the balance by correcting the perturbed body posture. For example, one of the consistent corrective movements involved bringing the left foot forward faster after experiencing a walkway perturbation. Moreover, bilateral cocontraction of the lumbar muscles was observed at the touch down (heel strike) phase during normal walking, slip, and trip trials, which supports prior literature reports (e.g., Potvin & O'Brien, 1998; Thorstensson, Carlson, Zomlefer, & Nilsson,1982).

The bilateral cocontraction of the lumbar muscles may increase after the heel strike directly after experiencing a slip or trip. Hence, in order to recover from a slip or trip event, the high muscle activation levels (including the period of lumbar bilateral cocontraction) may cause stiffening of the spine in order to provide stability to the lumbar spine to prevent injury (or fall). It is notable that start point of a participant's walk was adjusted so that the right foot experienced the perturbation. These substantial levels of muscle activity in awkward postures after the walk perturbations account for resulting low back pain - even in the absence of a subsequent fall.

4.3. Lumbosacral Reaction Forces and Moments

The current study used EMG-based musculoskeletal model to estimate the lumbosacral reaction loads using kinematics, muscle activity, and anthropometric information as the main inputs. The EMG-based lumbosacral model used in this study (Jia et al., 2011) represents a more refined model, which provides detailed muscle anatomy and incorporates muscular dynamics for predicting lumbosacral forces and moments.

The findings in the present investigation supports prior studies in terms of the lumbosacral loads during normal walking. For example, Cappozzo (1984) and Callaghan et al. (1999) reported the peak lumbosacral loads acting at the lumbar region in the range of 100-250% of the weight (BW) during normal human walking. Another study predicted the peak compressive loads during normal walking within a range of 92-345% of BW (Khoo et al., 1995). Callaghan and colleagues (1999) also reported the peak compression forces within the range of 46-204% of BW. For the current study, during normal walking the peak compressive forces and the peak resultant forces were noted to be 150-219% and 153-224%, respectively, of average body weight (ABW) across the different participants. In a similar study, the Khoo Group (1995) reported the A/P shear forces to be 22% of BW; in the current study, however, we noted the A/P shear forces to be slightly higher - namely 37.1 (8.9)% of ABW (SD). This discrepancy can be related to the differences in experimental details and the different modeling approach between the two studies. Khoo and coworkers (1995) used a non-EMG-based biomechanical model that calculated the lumbosacral forces and moments using force-plate data, motion capture data, and participants' anthropometric information as the main inputs. The lateral shear forces reported by Callaghan et al. (1999), who employed an EMG-based model, were in the range of 12-58% of BW, which corresponds well with the lateral shear forces reported in the current study: 11.6 (5.1)% of ABW (SD).

Of the tri-axial forces (lateral shear, A/P shear, and compression) reported in the current study, compressive forces remained dominant throughout normal walking, slip, and trip trials across all participants. We associate this outcome with the fact that during any specific task performed by humans (i.e. sitting, normal walking, running), the L5/S1 joint almost always remains under compression - principally due to the weight of the trunk and upper body imposed by gravity. The peaks in the loads were observed to occur during the single stance phase, wherein the lumbar spine is supported by a single limb. The peak loads produced during the gait cycle are believed to be indicative of the most adverse effects at the lumbosacral joint (Khoo et al., 1995). For the slip and trip trials, peak resultant lumbosacral forces were noted to achieve a near-threefold

Table 7: Summary of lumbosacral forces expressed as a percentage of the average body weight (ABW) (SD) across the different participants during normal walking, slip and trip trials.

Force (N)	**Normal Walking**	**Slip**	**Trip**
A/P Shear Force	37.1 (8.9)	149.5 (29.7)	128.3 (45.1)
Lateral Shear Force	11.6 (5.1)	25 (13)	35.4 (25.8)
Compression Force	184.6 (34.2)	446.9 (78.2)	445.1 (96.3)
Resultant Force	188.7 (35.3)	472.3 (81.7)	465.9 (103.5)

increase in the values to 472.3 (81.7)% for slip trials, and 465.9 (103.5)% of ABW (SD) for trip trials compared to normal walking. Table 7 provides a detailed summary of the individual and the resultant lumbosacral forces as a percentage of ABW during normal walking, slip and trip trials.

Table 8: Summary of lumbosacral moments expressed as a percentage of the average body weight times height (ABWH) (SD) across the different participants during normal walking, slip and trip trials.

Moment (Nm)	Normal Walking	Slip Trial	Trip Trial
Lateral Bending Moment	1.1 (0.5)	3 (1)	2.9 (1.2)
Flexion-Extension Moment	0.7 (0.3)	3.4 (1.8)	2.7 (1.3)
Axial Rotation Moment	0.5 (0.2)	1.4 (0.8)	2.1 (0.8)
Resultant Moment	1.5 (0.4)	4.9 (1.7)	4.7 (1.1)

The lumbosacral moment data obtained during normal walking in this investigation are consistent with values reported in the existing literature. Using an EMG driven model, for example, Callaghan et al. (1999) reported the peak lateral bend moments, flexion-extension moments, and axial rotation moments within the range of 0.31-4.44%, 0.62-2.87%, and 0.15-1.04% of body weight times height (BWH), respectively. The values for the respective moments in the current study were found to be 1.1 (0.5)%, 0.7 (0.3)%, and 0.5 (0.2)% of average body weight times average height (ABWH) (SD) across all the participants. Resultant lumbosacral moments during normal walking were reported to be 1.5 (0.4)% of ABWH. As previously observed with lumbosacral forces, the peak resultant lumbosacral moments during the slip and trip trials increased to more than a threefold value of 4.9 (1.7)% and 4.7 (1.1)% of ABWH (SD), respectively, compared to normal walking. A general overview of the individual and resultant lumbosacral moments expressed as a percentage of ABWH during normal walking, slip, and trip trials is represented in Table 8.

4.4. Low Back Pain Development

Previous studies have reported low back pain to be a potential outcome of slips and trips (Grönqvist et al., 2001; Manning & Shannon, 1981; Pope, 1989; Bentley & Haslam, 1998). However, the current study investigated the effects of slips and trips that do not lead to a fall, but nonetheless can potentially be hazardous to the spine because of the rapid corrective actions made by the body to regain balance. A significant body of epidemiological research supports the fact that unexpected and unanticipated movements made to regain balance can lead to the development of low back pain (e.g., Lavender et al., 1988; Stobbe, & Plummer, 1988; Manning et al., 1988; Manning, Mitchell, & Blanchfield, 1984; Rohrlich, Sadhu, Sebastian, & Ahn, 2014). The current investigation adds to the scholarship in this area by explaining why STWFs can be injurious to the lumbar spine. Indeed, overexertion in the lumbar region and related lumbar muscle activity demonstrated substantially large values to correct one's perturbed body posture due to a slip or trip. Subsequently, significant muscle activity resulted in substantial lumbosacral forces and moments during slip- and trip-induced trials compared to normal walking. Each of these response measures can be associated with low back pain development as follows:

- The additional lumbar motion during gait perturbations actually occurs due to a deviation of the body's COM (Center of Mass) subsequent to a slip or trip. This COM deviation can lead to awkward spinal postures - namely, non-neutral trunk postures in either extreme positions or angles (e.g., bending and twisting). The speed of change and degree of deviation from non-neutral posture is related to the risk of low-back injury (NIOSH, 1997) As documented in the current study, the awkward posture of the spine occurred very rapidly—roughly 250 ms after the heel strike on the force plate. It is presumed that a significant change in the motion of the TRP within such a short duration of time could lead to low back pain development (Marras et al., 1995).

- Similarly, the substantial increase in muscle activity that was documented during the slip and trip trials could lead to overexertion of muscles. The Bureau of Labor Statistics (2016) reported overexertion and resultant bodily reactions to be the one of the leading causes of injury. Accordingly, the overexertion of muscles could result in a low-back sprain, which could then exacerbate musculotendinous damage (Rashedi et al., 2012).
- Lumbosacral forces and moments were noted to undergo a substantial increase during the slip and trip trials compared to normal walking. For three of our participants, the peak compression force during a slip and trial exceeded NIOSH's (1981) permissible action limit of 3400 N. Similarly, the peak shear force reported during the slip and trip trials for all participants exceeded a proposed action limit of 500 N (McGill, Norman, Yingling, Wells, & Neumann, 1998). Such high force levels put the individual at risk for potential damage to the L5/S1 joint, thus leading to the development of chronic low-back pain.

All the response measures are, in fact, interrelated. After experiencing a walkway perturbation, one's lumbar kinematics will increase significantly within a very short period of time (approximately 250 ms after the heel contact on the force plate), resulting in the deviation of posture of the TRP. Moreover, cocontraction of the muscles observed during the slip and trip trials causes the lumbar spine to stiffen, likely to provide stability to the spine to prevent injury. Consequently, the activation level of the muscles increases significantly - due in part to correct the posture deviation. As a result, a significant increase in the lumbosacral forces and moments occurs due to the effort made by the body to regain balance after experiencing a slip or a trip.

5. Limitations & Future Work

A number of limitations should be noted for the current study. First, we were unable to use an inverse dynamics approach to validate the lumbosacral moments obtained from the EMG-based model. As discussed earlier in the Results section, the relatively poor quality of the ground reaction forces collected from the force platforms precluded the possibility of calculating moments using an inverse dynamics model. Using the bottom-up approach for inverse dynamics, these errors compound when moving further along the chain of segments, which intensified at each step since each segment solution depends on the reaction forces from the previous segment.

Another limitation in the current study that must be noted pertains to the sudden change in our experimental floor surface from a “normal” surface to a slippery surface or one featuring a hazard through the activation of a trip plate. Although the floor transition was performed without the participants’ knowledge, such conditions may or may not simulate an occupational slip or trip (or slips/trips that occur in the performance of ADLs). Indeed, during the consent process we had to declare that there would be some sort of gait perturbation. Despite our best efforts to divert their attention, there might have been some expectancy on the part of participants that a slip/trip-inducing condition was going to occur. This scenario introduces the possibility of heightened caution on their part to prevent themselves from falling - which would not be the case in an occupational setting. However, we did make a point of asking them about their expectancy of the gait perturbation right after its occurrence, and results did not show any serious concern in this regard.

This study is also limited by other potential source of inaccuracies inherent to the experimental data-collection process utilized herein. Consider, for example, the accuracy and repeatability of the marker positioning on the bony landmarks for different participants. There could have been the possibility that the soft tissues on the bony landmark enabled the marker to slide somewhat on the skin with respect to the bone. This slight shifting could have skewed the data due to differences in the rate of motion between the marker and the body segment.

The limitations associated with the present study lay a groundwork for future investigations. For example, in addressing one of the main limitations discussed above, a future study could conduct a full-body 3D inverse dynamics analysis to verify lumbosacral moments through the amassing of a broader set of experimental data. Moreover, the current study included a fairly young pool of participants (20-28 years). Hence, it would be interesting to extend the scope of the project and investigate the effects of STWFs using an experimental cohort of middle-aged and elderly people. While this study investigated the kinematics, muscle activity, forces and moments for the low back, a similar model could be developed to determine the kinematics, muscle activity, forces and moments on other body segments/joints (e.g., lower extremities). Despite the discussed limitations, the current model can be used to develop effective training regimes regarding the specific body mechanisms that could be adopted after experiencing a slip or trip. Such preventative strategies could help reduce and control the number of falls as a consequence to slips and trips.

6. Conclusion

The current study investigated the effects of slips and trips on lumbar kinematics, lumbar muscle activity, and lumbosacral forces and moments. An EMG-based lumbar spine model was used, along with the calculated kinematics and muscle activity data, to estimate the lumbosacral forces and moments. Results from the current study indicate that lumbar kinematics, lumbar muscle activity, and low-back loads increased significantly during unexpected and unanticipated walkway perturbations such as slips and trips compared to normal walking. The study indicated that one of the main reasons for such high loads on the spine could be due to the rapid corrective actions (large muscle activity) made by body to regain balance after experiencing a slip or a trip in order to prevent a fall. Such high load levels could be hazardous for the lumbar spine and could lead to the immediate or later onset of low-back pain. Results from the current study can be used to develop intervention techniques involving the control of specific mechanism related to low-back disorders in order to reduce the risk of slip- and trip-related injuries.

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